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THE POTENTIAL OF A TESLA TYPE DEVICE AS A NON PULSATILE BLOOD PUMP.

A thesis submitted to Middlesex University in partial
fulfilment of the requirements for the degree of PhD.

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December 2006.

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**THESIS
CONTAINS
CD/DVD**

Abstract.

A review of the published work on pumps designed to assist a failing biological heart has been made and discusses the different types of pump presently on the market with the characteristics of each. The materials used to make these pumps are also discussed, together with some of the methods of output control. The application to the patient is described together with the advantages and disadvantages.

A Tesla type pump appeared to offer an alternative solution to those problems listed above. This is not a new design but appeared to offer advantages if applied to the application of pumping blood from outside the body. One of these was that at a constant speed, the pump supplied constant fluid pressure irrespective of the delivery. It also appears that the pump can give fluid shear stress levels that are less than the amount that will seriously damage blood components.

A prototype pump has been built and tested. The pump achieved the performance target delivery of 10 l/min at a differential pressure of 200 mm Hg. This was considered to be greater than the average performance produced by existing blood pumps but the maximum performance that could be produced by the human heart under extreme conditions. The pump reached a maximum speed of about 4000 rev/min with a maximum power consumption of about 120 Watts.

The results indicate that this type of pump is a potential blood pump in terms of the delivery and pressures achieved.

The characteristic performance figures are within the envelope of published theoretical results.

The pump tested here needs further development to improve the hydraulic performance. Recommendations are made for the direction of future work to improve the pump efficiency and flow patterns, biocompatibility and methods of production. Controls and power supply also need improvement.

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Glossary

Aggregate. A dense cluster of cells wherein macromolecules such as fibrinogen physically link adjacent cell membranes.

Aggregation. A crowded mass of independent but similar units, a cluster.

Albumin. A water soluble protein coagulated by heat. Serum albumin is found in blood plasma and is important for the maintenance of the blood plasma. Albumin is synthesized in the liver and the inability to do this is a feature of chronic liver disease, i.e. cirrhosis. The type of albumin depends on the source, muscle, milk, blood, serum, egg, vegetable etc. Some albumins, (egg), injected directly into the blood are highly poisonous.

Allograft. A transplant between individuals of the same species, i.e., kidney transplant between two unrelated humans.

Angina. A sense of suffocation or suffocating pain with difficulty in breathing or choking. This term is often used to describe Angina Pectoris. It is usually a pain in the centre of the chest induced by exercise and is relieved by rest. In this condition the heart is demanding more blood than can be supplied through the coronary arteries, probably due to artery disease. The condition can be relieved by drug treatments or by invasive surgery.

Anomalous. Any deviation from normal. Something that is irregular or deviating from rule.

Antibody. A special type of blood protein synthesized in lymphoid tissue in response to the presence of a particular antigen, which it attacks and renders harmless. The antigen can be contained in pollen grains, invading bacteria, foreign blood cells, infectious diseases and can be responsible for organ rejection.

Anticoagulants. Drugs which prevent the coagulation of blood. The main ones in use: - heparin, phenindione, and warfarin.

Antigen. Any substance that the body regards as foreign or potentially dangerous and against which it produces an antibody. Antigens can trigger the immune response following organ transplantation.

Apical. The apex or tip or summit of an organ.

Apparent viscosity. The viscosity calculated from Poiseuille's law at any particular flow and tube diameter; it is used for suspensions such as blood, that exhibit anomalous viscosity, the ratio of shear stress to shear rate at any point in a viscometric flow of an inhomogeneous fluid.

Arrhythmia. Any variation from the normal regular rhythm of the heart beat. Can be a symptom of heart disease but can also occur without apparent cause.

Artery. A blood vessel carrying blood from the heart to an organ(s). The walls of an artery consist of four layers, a smooth inner lining, a thin elastic layer, a thick muscular, elastic, layer and a tough fibrous outer layer. The structure of arteries enables them to withstand the high blood pressure peaks caused by pulsatile action of the heart. See also veins.

Glossary, (continued).

Aspirin. (Acetylsalicylic acid.) Widely used drug used for relief of pain, inflammation and fever. Taken by mouth, alone or in combination either other analgesics for the relief of the less severe types of pain.

Daily doses used to help prevent coronary thromboses / strokes in those at risk and have a protective effect against other conditions

Atheroma. Degeneration of artery walls due to the formation of fatty plaques and scar tissue.

Atherosclerosis. Disease of the arteries where fatty plaques develop on the walls of arteries and can cause blockage to blood flow. See atheroma.

Autograft. A transplant within one individual, from one part of the body to another. A skin graft would come into this category.

Autologous – vein. Denoting a graft that is derived from the recipient of the graft. Can also indicate a natural 'by-pass' of a blocked artery/vein. (i.e., Multiple capillaries by-passing a blockage.)

Azotaemia. The presence of urea and other nitrogenous bodies in greater concentrations than normal in the blood. Usually associated with advanced types of kidney disease. (Former name for uraemia, occurs in kidney failure).

Bilirubin. The chief pigment in human bile, derived from haemoglobin.

Bioactive material. A material designed to promote a specific response such as bone in-growth.

Biocompatible. A material able to perform with a host in a specific situation.

Bioinert material. A material which causes little response from the body reactions.

Biomaterial. Any substance, other than a drug, or combination of substances, synthetic or natural in origin, which can be used for a period of time, as whole or part of a system, which treats, augments, or replaces any tissue, organ, or function of the body. (Boretos, J.W. and Eden, M, Eds., Contemporary Biomaterials, Park Ridge, New Jersey. Noyes Publications, p651 – 665. (1984)

Blood pressure. This is the pressure of the blood produced by the heart. Blood pressure varies with each patient and is related to age and physical condition. Accepted figures for a healthy adult are a systolic pressure in the order of 140 mm Hg. and a diastolic pressure of 80 mm Hg.

Cannula. A hollow tube designed for insertion into a body cavity, (bladder / blood vessel).

Carbon dioxide - can poison haemoglobin. The blood of cigarette smokers contains more carbon dioxide than normal and babies born to smoking mothers are usually smaller due to less oxygen received during fetal development.

Glossary, (continued)

Cardiomyopathy. Diseases of the heart muscle, origin unknown, any chronic disorder affecting the heart muscle, can be inherited but can also be caused by other conditions, viral infections, alcoholism, vitamin B deficiency, (beri-beri), and amyloidosis, cause often unknown. May result in enlargement of the heart, heart failure, arrhythmias and embolism. Often no specific treatment but patient often improves following control of heart failure and arrhythmias.

CFD. Computational Fluid Dynamics, a mathematical calculation used to determine the flow patterns in fluids.

Cholesterol.

Chemically a lipid, cholesterol is an important constituent of body cells. It is involved in the formation of hormones and bile salts and in the transport of fats in the bloodstream to tissues throughout the body. There is evidence that a high blood cholesterol level increases the risk of atherosclerosis, (accumulation of fatty deposits on the lining of arteries), and the risk of coronary artery disease or stroke.

Clotting factors. A group of substances present in blood plasma that under certain circumstances undergo chemical reactions leading to the blood clotting process. Although they have specific names most are referred to by an agreed set of Roman numerals. Lack of any of these factors results in the inability of the blood to clot.

Collagen. A protein, principal constituent of white fibrous connective tissue, occurs in tendons. Also found in skin, bone, cartilage and ligaments. Relatively inelastic but has high tensile strength. An important constituent of heart and blood vessels and with calcium salts provides the rigid structure of bone.

Covalent bond. A chemical bond in which electrons are shared between two atoms, giving each atom a share in each other's electrons.

Dipyridamole. A drug that dilates the blood vessels of the heart, reduces platelet aggregation. Given by mouth or injection to prevent thrombosis.

Embolus. Any material that lodges in a vein or artery.

Embolism. The blocking of a blood vessel by an embolus.

Endothelial cells. Endo – prefix meaning located inside.

Endothelium – membrane lining various vessels and cavities of the body – fibrous layer covered with thin, flat cells, renders the surface perfectly smooth, secretes the fluid for lubrication

Erythrocyte. Another name for the red blood corpuscle. A blood cell containing the red pigment, haemoglobin, the principle function of which is to transport oxygen around the body.

Extracorporeal. Outside or unrelated to the body.

Glossary, (Continued)

Factor. A substance that is essential to a physiological process – often a substance the nature of which is unknown.

Factor XI. A coagulation factor normally present in blood. Deficiency of the factor is inherited but rarely causes spontaneous bleeding. However bleeding does occur after surgery or trauma to the blood vessels.

Factor VII. A single chain glycoprotein that is essential for the process of coagulation / formation of blood clots.

Fibrillation. A rapid and chaotic beating of the many individual muscle fibres of the heart, the consequence of this is that the heart is unable to maintain effective synchronous contractions.

Atrial fibrillation, a common type of arrhythmia, a rapid and irregular heart rate.

Ventricular fibrillation, occurs when the heart stops beating, a cardiac arrest, and is the cessation of effective pumping of the heart where the muscle fibres are beating rapidly without pumping blood.

Fibrin. Substance formed in the blood as it clots. Produced as threads forming a close meshwork through the blood, formed not only from shed blood but also from lymph via lymph vessels. Precursor is fibrogen, soluble protein in blood.

Giant cells. Any large cell such as a megakaryocyte – may have 1 or many nuclei.

Mega – prefix denoting largeness

Megokaryocyte – a cell in the bone marrow, (35 - 160µm in dia.), that produces platelets.

Glutaraldehyde. A chemical used as a disinfectant in health care and other industries. It can cause irritation to human skin in concentrations above 2%. Extended exposure has caused skin dermatitis and may cause asthma.

Haemocompatibility. The capability of a substance or material of mixing with and existing with blood with no adverse effects on the function of each other.

Haemodialysis. The technique of removing waste products and poisons from the blood by what is basically a filtering process.

Haemoglobin. The material that produces the red colour of blood. It is a dynamic molecule that changes its shape and characteristics and helps regulate oxygen delivery and blood pH levels. Main function is to carry oxygen around the body. It binds to oxygen when exposed to air, transports oxygen to capillaries, releases oxygen, binds to carbon dioxide, and then transports the carbon dioxide to the lungs

Haemolysis. The breaking up of the blood corpuscles by the action of poisonous substances, usually of a protein nature, circulating in the blood or by certain chemicals. It can occur gradually in some forms of anaemia and rapidly in poisoning for example via snake venom. The destruction of red blood cells.

Glossary, (Continued)

Haemostasis. The arrest of bleeding involving the coagulation of blood / contraction of damaged blood vessels. Can include the application of ligatures or diathermy.

Haematocrit. The proportion of red blood cells in a column of centrifuged blood + plasma mixture. Normal figures are plasma = 55%, red blood cells = 45%.

Heart output. At rest, output is approx. 6 litres/minute @ 60/80 beats/minute. Active output can be as high as 50 litres/minute @ 200 beats/minute.

Heparin. An anti-coagulant produced by the liver, some white cells and other sites such as muscle and lung. It inhibits the action of the enzyme thrombin in the final stages of blood coagulation. An extracted, purified form is used for the prevention of blood coagulation in patients with thrombosis and similar conditions. It is also used in blood collected for examination.

Complicated in chemical structure, has been prepared in crystalline form. Carbohydrate in nature, contains amino and sulphuric acid groups.

Usually given by injection, not effective when given by mouth. The most important side effect is bleeding.

See also 'low molecular weight heparin' and 'anticoagulants'.

Heterograft. A transplant between members of a different species, i.e., heart valves taken from a pig and transplanted into a human.

Homograft. See Allograft.

Hydrophilic. A surface having affinity for water, 'wetable' by water.

Hydrophobic. A surface not readily 'wetable' by water.

Hyperplasia. Abnormal multiplication or increase in the number of normal cells in a normal arrangement of tissue.

Hypertrophy. This is the increase in the size of a tissue or organ by the enlargement of its cells rather than by its multiplication of cells as in normal growth. Examples are the enlargement of the breast during pregnancy, enlargement of remaining kidney following failure of the first.

In vitro. Latin, meaning 'in glass', referring to a laboratory test.

Intima. Innermost coat, lining the arteries and veins.

Inotropic. A substance affecting the contraction of heart muscle. Digitalis, dobutamine, enoximone have a positive action stimulating contractions and the heart rate to increase. Betablockers such as propranolol have a negative reaction reducing contractions and the heart rate to decrease.

In vivo. Latin meaning 'in life', referring to tests in living systems.

Glossary, (Continued)

Ischaemia. Bloodlessness of a part of the body due to contraction, spasm or blocking, (by embolus / thrombus).

Isograft. A transplant between two genetically identical individuals, i.e., identical twins.

Isomer. The same kind and number of atoms united in different structural forms giving different compounds with different characteristics. Similar to allotropism or polymorphism in crystalline materials where the same material can exist in more than one crystal structure.

Laminar flow. The relative motion of elements of a fluid along smooth, parallel paths, this occurs at lower values of Reynolds numbers.

Leucocyte. A white blood cell. Any blood cell that contains a nucleus.

In health there are 3 major subdivisions, granulocytes, lymphocytes and monocytes that are involved in protecting the body against foreign substances and in antibody production. In disease a variety of other types may appear in the blood most notably immature forms of the normal red or white blood cells.

Leucocytes constitute one of the most important of the defence mechanisms against infection, particularly neutrophil leucocytes.

Ligature. A cord or thread tied around an artery to stop the blood flow or to prevent the escape of blood from a cut end. The cord is usually cat gut or silk and tied with a reef knot.

Lipid.

A general term for fats and oils. Lipids, or lipins, as they are sometimes called include triglycerides, (simple fats), phospholipids, (important constituents of cell membranes and nerve tissue), and sterols such as cholesterol.

Low molecular weight heparin. A type of heparin more readily absorbed and requires less frequent administration than standard heparin. Used as anticoagulant to prevent deep vein thrombosis after surgery/kidney dialysis. Less risk of bleeding. Also known as 'enoxaparin'.

Lysis. The gradual ending of a fever, also used to describe the process of dissolution of a blood clot / loosening of adhesions. Denotes dissolution.

Macrophage. A large scavenger cell, (phagocyte), present in connective tissue.

Fixed cell. Stationary within connective tissue / major organs / bone marrow / nervous system.

Free cells wander between cells and aggregate at focal sites of infection removing bacteria / foreign bodies from blood and tissue.

Mediastinum. The space in the chest between the lungs that contains the heart, great vessels, gullet, lower part of windpipe, thoracic duct, phrenic nerves and other structures. Regions are anterior, middle, posterior and superior.

Morphology. This is the study of forms of things, especially of animals, and their structures.

Glossary, (Continued)

Myocarditis. Chronic or acute inflammation of the heart muscle. May be seen alone or as part of pancarditis when the pericardium is involved.

Newtonian fluid. A fluid in which flow and shear rate are always proportional to the applied stress; a fluid in which the ratio of shear stress to shear rate in a viscometric flow, defined as the dynamic viscosity, is independent of shear rate and is at most a function of temperature.

Non – newtonian fluid. A fluid in which flow and rate of shear are not proportional to the applied stress; a fluid in which the ratio of shear stress to shear rate in a viscometric flow is a function of shear rate as well as temperature.

Oedema. An abnormal accumulation of fluid in the body tissues. Local – from injury giving local swelling, or general – causes heart failure, kidney or liver problems.

Orthotopic. Orthotics – science and practice of fitting surgical appliances to assist weakened joints.

Pacemaker. A device used to produce and maintain a normal heartbeat in patients who have heart block, (sinoatrial node). It consists of a battery and electronics that stimulate the heart via. a wire attached to the surface of a ventricle, (epicardial pacemaker), or lying in contact with the lining of the heart, (endocardial pacemaker). Some pacemakers are temporary, others are permanent, some stimulate at a fixed rate, others sense when the heart rate falls below a pre-determined rate which then triggers the device, (demand pacemaker).

Pannus. General name given to additional, unwanted tissue growth.

Perfusion. 1) The passage of a fluid through a tissue, especially the passage of blood through lung tissue to pick up oxygen from the air in the alveoli, which is brought there by ventilation and release carbon dioxide. If ventilation is impaired, deoxygenated venous blood is returned to the general circulation. If perfusion is impaired, insufficient gas exchange takes place.

2) This is the deliberate introduction of a fluid into a tissue, usually by injection into the blood vessels supplying the tissue.

Pericardium. Sac or membrane lining the heart.

Phagocyte. A defensive cell that is able to release powerful chemicals to kill invading cells, will engulf and digest bacteria, protozoa, cells and cell debris and other small particles. Includes White blood cells, (Leucocyte), and Macrophages. Phagocytes take up oxidised low-density lipoprotein, (the major cholesterol carrier in the blood), become 'foam' cells which congregate on arterial walls. These events are believed to participate in the build up of lipid deposits leading to clogged arteries.

Phagocytosis. Process by which the attacks of bacteria upon the living body are repelled and the bacteria destroyed via. the action of the white corpuscles of the blood.

Glossary, (Continued)

Poly. Meaning 'many'.

Polymer. Chemical compound with large molecules made of many smaller molecules of the same kind, i.e. a molecule comprising of 'many mers'.

Polymorph. Any one of several forms in which the same thing may occur. An organism occurring in several forms, / a substance crystallising in several systems.

Pulmonary capillary wedge pressure. (PCWP) The pressure of blood in the left atrium, indicating the adequacy of the pulmonary circulation.

Pyrolytic carbon. Pyro – prefix meaning anything connected with fire or produced by heating. A biocompatible material, can be used as a coating.

Rejection. Process where the body rejects a 'foreign' body. This can be a valuable part of the body defence mechanism but can act adversely if such organs as heart and kidneys are the implanted body as the rejection can lead to the death of the patient. See Tissue Typing.

Reynolds number. A dimensionless number that is used to describe the flow conditions of a fluid in terms of density, viscosity and velocity in a tube or other flow passage.

Rheology. The science of the deformation and flow of matter.

Rouleaux. An aggregate of erythrocytes stacked like a pile of coins.

Shear rate. The change in velocity of a flowing liquid separated by unit distance; its units are expressed in inverse seconds, (s^{-1}). The velocity gradient of fluid layers in a given vessel; when flow velocity is high, shear rate is high, and vice versa.

Shear stress. Is a measure of the force required to produce a given rate of flow and is expressed in N/m^2 .

Sinoatrial node. The pacemaker of the heart, a microscopic area of specialised cardiac muscle located in the upper wall of the right atrium near the entry of the vena cava. Fibres of the sinoatrial node self excite and contract rhythmically around 70 times/min. Following each contraction, the impulse spreads through the atrial muscle and into fibres connecting the sinoatrial node with the atrioventricular node. Fibres of the autonomic nervous system supply the sinoatrial node and the impulses arriving at the node either accelerate or decelerate the heart rate.

Stenosis. The abnormal narrowing of a passage or opening such as a blood vessel or heart valve.

Sternotomy. Surgical division of the breast bone, (sternum), to allow access to the heart and major vessels.

Glossary, (Continued)

Systole. The period of the cardiac cycle during which the heart contracts. The term usually refers to ventricular systole, which lasts about about 0.3 seconds. Atrial systole lasts about 0.1 secs.

Tachycardia. An increase in the heart rate above normal. Can be caused by exercise – excitement – illness.

Tampon. A pack of gauze, cotton wool or other absorbent material used to plug a cavity or canal in order to absorb blood or secretions.

Tamponade. Abnormal pressure on a part of the body, for example, pressure caused by the presence of excessive fluid between the pericardium, (sac surrounding the heart), and the heart.

Tetralogy of Fallot. Cyanotic congenital heart disease. (The child is blue, i.e., a 'blue baby'.) This can be a) Stenosis of the pulmonary valve, b) Septal defect, (ventricular defect), c) Aorta over rides both ventricles, d) Marked hypertrophy of the right ventricle.

TETS. Transmission of Energy Through the Skin. A process where electrical power is transmitted through the skin of the patient and is used to charge an implanted battery. The implanted battery is then used to power an implanted control system and blood pump.

Thrombus. Blood clot or the formation of blood clots.

Thrombogenic. An agent which promotes the development of a thrombus.

t-PA. (Tissue plasminogen activator.) A natural protein in the body that can break up a thrombosis. It can now be produced by genetic engineering. It requires the presence of fibrin as a co-factor and is able to activate plasminogen on the fibrin surface which distinguishes it from the other plasminogen activators, streptokinase and urokinase.

Tissue Typing. This process compares the compatibility of different tissues to the immunological responses and is known collectively as histocompatibility. Equality, or near equality, of tissue types is essential to ensure acceptance of grafts, transfusions and transplants.

Tribology. The science and technology embracing all subjects involved when surfaces in contact move in relation to one another.

Vein. A blood vessel carrying blood away from an organ(s). A vein consists of three layers, an inner coating, a muscular middle layer, and a tough outer layer. The walls of veins are thinner than the walls of arteries and most contain valves to ensure that blood flow is in one direction only. Veins do not have the pulsatile blood flow associated with arteries. See also arteries.

Viscosity. In general, the resistance to flow or alteration of shape by any substance as a result of molecular cohesion; most frequently applied to liquids as the resistance of a fluid to flow because of a shearing force.

Glossary, (Continued)

Warfarin. Anticoagulant used in treatment of coronary and venous thrombosis to reduce the risk of embolism. Usually given by mouth, maximum effect in 36 hours, action passes off within 48 hours of cessation of treatment. Principal toxic effect is bleeding usually from gums and other mucous membranes.

White blood cell. Any cell that contains a nucleus. Three major sub divisions, granulocytes, lymphocytes, monocytes – all involved in protecting the body against foreign substances and in anti body production. See leukocyte.

Xenograft. See Hetrograft.

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CHAPTER 1, INTRODUCTION.

1.0 Aims

The purpose of the present work is to investigate the design and parameter measurements of a Tesla type pump, (Tesla N., 1856-1943), to assess its suitability to pump blood in vivo.

In order to understand the constraints involved in pumping blood a review of the biological and physiological problems involved including effects on the heart and other organs is made.

1.1. The Heart.

The left ventricle of the heart pumps blood through arteries to all the organs in the body where it returns through veins to the right atrium to be pumped around again. The failure to pump sufficient blood is linked to cardiovascular problems. Signs of heart failure include fatigue, breathlessness after exertion and the accumulation of fluid in the lungs, ankles and abdomen. The inability of the heart to pump enough blood can cause damage to organs such as the kidneys, liver, which can result in death from organ failure in addition to any heart problem.

Initial treatments include drug therapy and lifestyle changes. If treatment, such as angioplasty, is not satisfactory, the final solution is a heart transplant. This procedure carries risks in terms of rejection. Sufficient donor hearts are difficult to obtain and the supply in the USA in the years from 1992 to 2002 has remained at about 2500 per year with a waiting list of up to 8000 potential recipients. Zaroff J. et al, (Circulation 2002). Many patients die before receiving a transplant.

Alternative treatments include the use of mechanical aids to pump the blood around the body. These devices assist or replace the heart and each one carries its own set of problems and risks. (New and Emerging Techniques – Surgical, 2002)

Table 1.1 shows the range of possible therapeutic options using mechanical pumps to pump blood.

Therapeutic options	Description
Bridge to transplant	To support the heart until a transplant can be performed.
Bridge to recovery	To be used temporarily while the heart recovers from heart failure.
Post-cardiotomy recovery	To provide circulatory support to a patient after open-heart surgery.
Destination therapy	To provide permanent support to the heart, replacing the need for a transplant.

Emerging Growth capital Pty Ltd., 2004

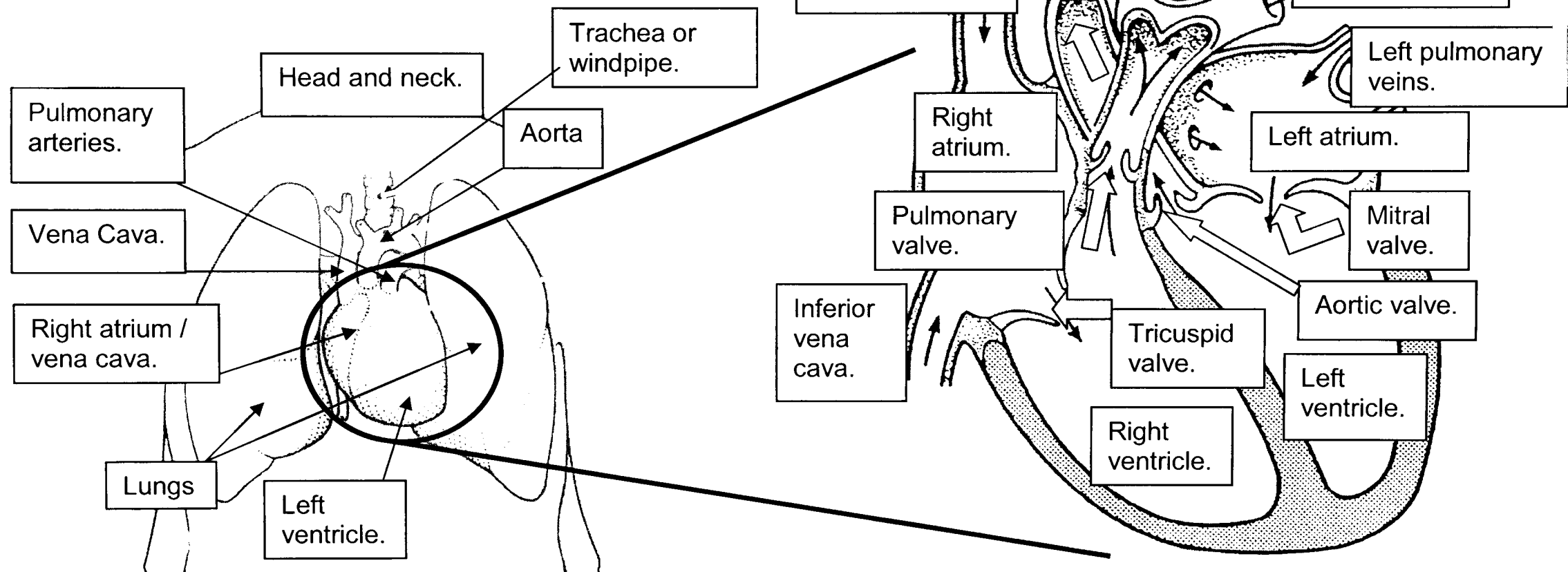
Table 1.1, Therapeutic options for LVAS, (Left Ventricular Assist Systems)

1.1.1 Description of the human heart.

The adult human heart is a hollow muscular organ of a conical form, placed between the lungs, and enclosed in the pericardium. The average heart measures 120 mm in length, 85 mm in the broadest part and 60 mm in thickness. The male heart weighs between 280 -340 g, and the female heart 230-280 g. (Grays Anatomy 1995). Figure 1.1 shows the position of the heart in the chest and the construction of the heart.

The heart is just left of centre in the chest and beats approximately 72 times a minute at rest, pumping around 5.5 litres/minute of blood for a man weighing 70 kg. The heart can increase output to over 11 litres per minute if required.

Figure 1.1. Position of the heart in the chest and the construction of the heart.



After: - BMA Complete Family Health Encyclopaedia, (1995)

From: - The Concepts of Intra-Aortic Balloon Pumping. Datascope Medical Co. Ltd

1.2 Heart problems and treatments.

Heart problems appear as cardiovascular events such as angina or heart attack.

Where the coronary arteries have a reduced flow, the heart muscle is not supplied with enough blood and the patient may experience pain known as angina during normal activities. Some relief can be given with rest or drugs, but surgical intervention is sometimes necessary along with control of lifestyle.

Heart attacks occur when any of the coronary arteries supplying blood to the heart muscle become blocked and the heart muscle is starved of oxygen and nutrients causing permanent damage to the muscle. Immediate medical treatment is necessary to relieve the condition. A serious attack may require the replacement of the heart depending on the amount of muscle damage.

With the heart functioning properly, the atria and ventricles beat in a synchronised sequence. If this does not happen, the ventricles will not pump blood effectively. A small electric shock, applied using a pacemaker, can keep the ventricle(s) running in phase.

1.3 Blood.

Blood can be considered as a suspension of deformable particles circulating within a system of flexible vessels. The deformable particles are the cells and Appendix 1 shows the characteristics of these. Each cell has its own function, for example, red cells transport O_2 and CO_2 and platelets are responsible for blood clotting. The flexible vessels are the arteries, veins and capillaries of the blood circulation system. Appendix 2 gives the dimensions of the vessels and the physiological flow parameters of the blood circulating through the system.

The blood has nutrients dissolved in it which are used to supply the muscles and organs with the 'fuel' they need to operate and it is the only part of the body that contacts every other part.

1.3.1 Blood as a fluid.

From Massey, (1995), where the relationship of shear stress and rate of shear is linear, the fluid is a 'perfect', or Newtonian fluid. The viscosity does not change with shear rate. With blood the relationship of shear stress and rate of shear is not linear and it is therefore a non – Newtonian fluid. The apparent viscosity changes with the shear rate.

If blood is made to flow through a tube, the rate of blood flow is not proportional to the pressure gradient and the apparent viscosity of blood falls as the pressure and flow rates rise. There is also an effect on apparent viscosity depending on the size of tube used. (Bayliss, 1952).

Under normal conditions, flow through a blood vessel does not result in adverse effects on the blood or blood vessel walls.

Cells that line the blood vessel walls are constantly exposed to mechanical forces generated by blood flow but changes to the fluid stress can modify the structure and function of these walls. This can lead to platelet growth and changes to the structure of the cells. (Chien, et al, 1998 and Resnick, Gimbrone, 1995.)

1.3.2 Red blood cells.

The primary function of red blood cells is to transport oxygen through the body. A red blood cell is a biconcave disc. It has a diameter of about 8 microns with maximum and minimum thickness of 2.4 and 1.0 microns respectively. The mean cell volume is about 87 cubic microns. There are about 4.5 to 5 million red blood cells per cubic millimetre of blood.

The membrane surrounding the cell allows water and ions to pass into the interior of the cell. It is possible for the cell to enlarge or shrink. and, within limits, these actions are reversible but if the limit is exceeded the cell will rupture and the contents will escape. The cell then dies, a process known as haemolysis.

A red blood cell does not renew or repair itself and under normal conditions it is worn out and will disintegrate after about 120 days. This releases the haemoglobin, which is broken down and reused where possible. The iron part of the haemoglobin can be returned to the bone marrow or can

be stored for future use. The rest of the cell is broken down and consumed by phagocytes. The normal destruction rate is around 3 million / second but adverse conditions can increase this rate. The body can increase output but reaches a maximum at 6 – 8 times normal but if requirements exceed this level the patient tends toward anaemia. (Harris, J.W., 1963).

1.3.3 Flow damage in blood.

Red blood cells can be destroyed by mechanical means. Abkarian and Viallat, (2005), report that when a shear is applied, the cells begin to rotate. The shape of the cell will be deformed from the normal bi-concave shape to more like an ellipsoid and the membrane is repeatedly stretched in a 'tank treading' motion as the cell rotates. The membrane eventually fails due to fatigue and the contents, (haemoglobin), are released into the plasma.

Damage to red blood cells is not necessarily instantaneous. Tests on dogs by Bernstein et al., (1967), indicated that immediate damage is a small proportion of the damage to blood during the shearing process. Their results indicated that there is an increasing lack of red blood cells as time progressed, implying the rate of cell destruction was greater than the rate of regeneration.

Rand, (1964), proposed that red cell damage occurred when the strain on the cell membrane exceeded a critical value whereafter the cell pores opened or the membrane ruptured releasing haemoglobin.

Work by Sallam and Hwang (1984) and Pinotti, (2000), indicate that eddies contained in turbulent flow could destroy red blood cells and that the destruction was related to the size of the eddy. They explained that if the smallest eddies in a blood sample are larger than the blood cell, any cells will be carried along within the eddy and will only experience stresses due to relative motion within the turbulent fluid itself. This condition indicates that any cell damage is dependant on time as the stresses are likely to be 'low'.

If the smallest eddies are the same size, or smaller, than the blood cell, the energy in those eddies will be transferred to the cell membrane. This may rupture, allowing the escape of haemoglobin if the eddies are powerful

enough. This condition indicates that cell damage is not only dependant on time but also on the energy contained in the fluid eddies.

Pinotti, (2000), developed the formula below to calculate the size of the smallest eddies.

$$L_s = \left\{ \frac{V^3 L_e}{U_o^3} \right\}^{0.25}$$

where:-

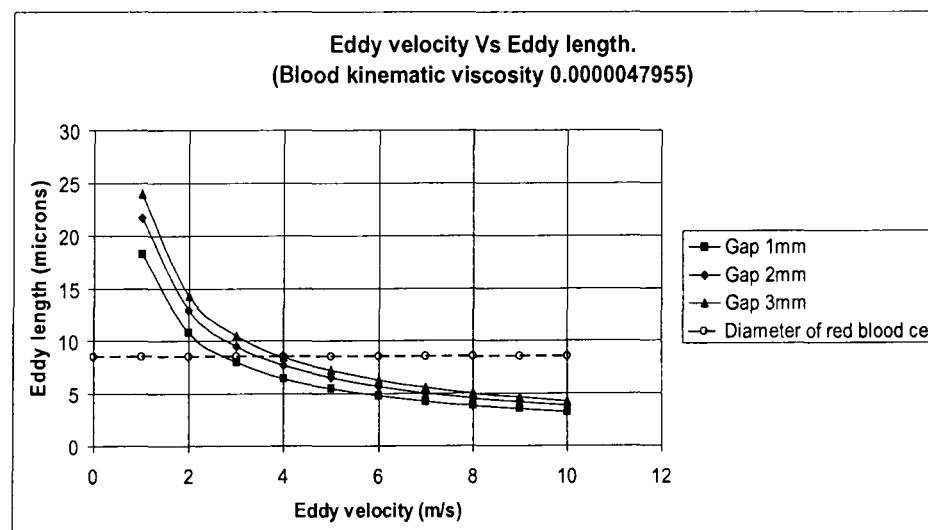
L_s = the calculated dimension of the smallest eddy.

V = the fluid kinetic viscosity

L_e = the characteristic length of the flow, for example, the diameter of a tube, (m)

U_o = the local RMS velocity (m/s)

Assuming in a radial flow blood pump application that L_e is the gap between the rotating disc and the volute wall of a rotary blood pump and that values of U_o range from 1 to 10 m.s^{-1} , using the formula, Figure 1.2 can be generated. The result indicates for gaps between 1 and 3 mm, any eddy velocity above about 3 m/s gives an eddy smaller than the cell and is likely to cause stress damage. Any velocity less than 3 m/s, the eddy length is greater than the red cell, maximum diameter 8 microns, which will then be carried along with the eddy and may not experience significant stresses. In all cases, the larger the gap, the greater the eddy length for a given velocity.



Calculated from Pinotti M., (2000)

Figure 1.2 Eddy velocity Vs Eddy length for gaps 1 to 3mm

It is important that fluid shear stress levels must be kept within acceptable levels. Nygaard and Paulsen et al., (1992), found the stress level just after a normal aortic valve was around 4 N/m^2 . With a partially blocked normal valve, a stress of 38 N/m^2 was recorded. With an artificial valve, stresses of up to 120 N/m^2 were found and they considered that this stress level was liable to damage blood.

Prec and Katz et al, (1949), calculated Reynolds numbers for the arterial system and suggested that there may be turbulence at the root of the aorta and in the neighbourhood of arterial branches.

Stein and Sabbah, (1976), measured the Reynolds number in the ascending aorta above the aortic valve in 6 normal subjects. They discovered Reynolds numbers of 5,700 to 8,900 and, during ejection, blood flow was '*highly disturbed*'. Flow through diseased valves and artificial valves produced turbulent flow with Reynolds numbers up to 10,000.

Goldsmith and Turitto, (1986), produced results showing high Reynolds numbers close to the aorta that could be interpreted as turbulent flow. The results, on Appendix 2, also showed low Reynolds numbers in the capillaries which would indicate laminar flow.

As turbulent flow is associated with high stresses, it is preferable that any blood pump produces laminar or non turbulent flow.

Blood damage.

There is little published information on exactly how much damage blood can withstand. If red blood cells are ruptured, the haemoglobin can escape into the plasma along with the cell fragments. Haemoglobin in the plasma can cause kidney damage. (Haemoglobin – Wikipedia. May 2006)

Leverett and Hellums et al., (1972), evaluating blood damage using a rotational centrifuge, suggested six factors liable to cause damage to red blood cells.

a) Solid - surface interactions.

Following observations by Blackshear, (1971), interactions with solid surfaces can damage red blood cells. The interaction will not necessarily

cause mechanical damage but could be cell - surface reactions leading to the formation of blood clots.

b) Centrifugal forces.

The centrifugal forces generated in a cylindrical viscometer tend to push the blood cells to the outer surfaces of the cylinder. Leverett, Hellums, et al., (1972), indicated this could result in cell damage independent of the effect of shear stresses. The blood will contact the internal walls of both radial and axial flow pumps as it passes through and damage to the blood may occur. The design of a blood pump will have to minimise this effect by reducing the centrifugal forces by using low rotational speeds and small internal diameters.

c) Haemolysis at an air – blood interface.

When blood is exposed to air, it coagulates and forms clots. There must be no blood – air interface inside the pump and this effect must not occur. Air bubbles in the circulating blood are dangerous as they could cause blockage to blood vessels with subsequent damage to the patient.

d) Cell – cell interaction.

Leverett and Hellums, (1972), estimated the haemolysis of red blood cells at fixed shear rates but at various hematocrit levels, (hematocrit is defined as the proportion of red blood cells in a sample of blood), to determine if the shear stress acted on individual cells or caused cell – cell interactions. They found that varying the hematocrit did not change the percentage of red cells haemolysed and concluded that cell – cell interaction was not an important mechanism in the destruction of red blood cells in the concentric cylinder viscometer. It is assumed that similar conditions apply inside a blood pump under the similar kinematic conditions.

e) Viscous heating.

During the operation of a blood pump, haemolysis may be caused by a rise in blood temperature. Tests by Leverett and Hellums, (1972), using a concentric cylinder viscometer and a shear rate of about of 300 N/m^2 for a 2 minute period indicated that a blood temperature rise of 10°C was possible. It

was concluded that if the blood temperature reached approximately 49°C. *'it will cause significant haemolysis and fragmentation'*.

Similarly Rosenberg, (1995), indicates that any device in contact with blood or tissue should not have a temperature rise of 5°C above the core temperature on a long term basis. For a normal human core temperature of approximately 37°C., this indicates a device maximum permissible temperature of 42°C.

f) Exposure time.

The effect of the exposure time of blood to surfaces and stress levels are important parameters governing blood haemolysis. It is not possible to give a 'limiting' figure for either time or stress as separate entities, as damage tends to be a cumulative effect. Leverett and Hellums, (1972), reviewed the results of prior work and indicated that there are two distinct regions affecting the threshold damage relative to time, one is shear stress and the other is surface effects. This relationship is shown in Table 1.4. It should be noted that not all the results reviewed are obtained with human blood and so may not be strictly applicable to human conditions but the data should be good enough for comparative purposes.

1.3.4 Stress in blood cells.

The amount of stress that a red blood cell can withstand before damage is difficult to determine. Table 1.2 shows some of the results obtained from various researchers.

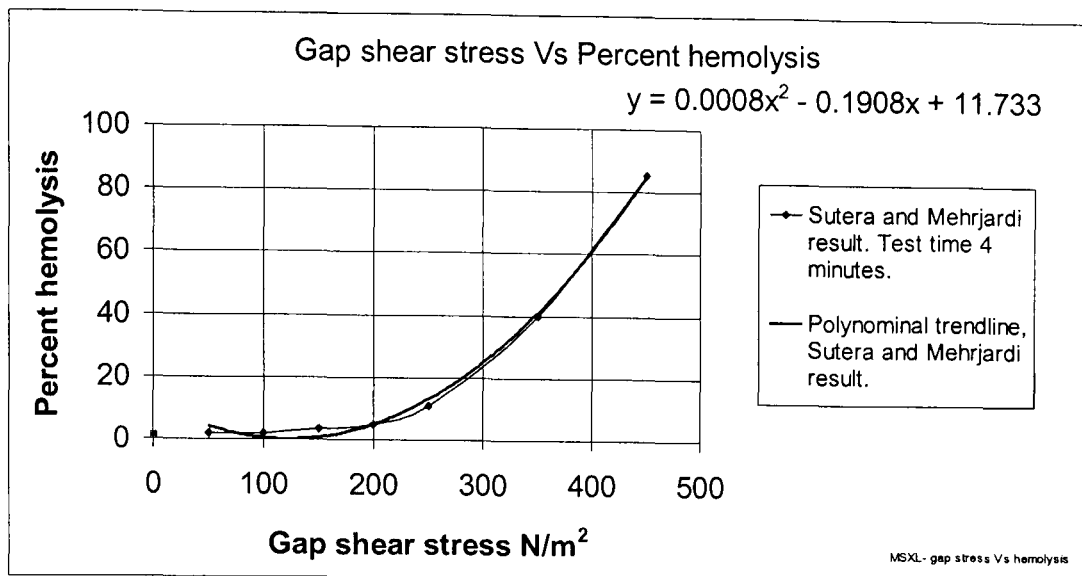
Stress N/m ²	Comments on cell condition.	Reference
4000	Stress level before damage	Bernstein et al, (1967)
150/400	Red cell destruction	Nevaril et al., (1969),
100/200	Bulk, medium stress, long exposure	Blackshear, (1972)
4000	Bulk, high stress, instant rupture	Blackshear, (1972)
70	Stress level before damage	Williams, (1973)
1/10	Stress level before damage, adjacent to foreign surfaces	Mohandas et al., (1974)
50	Cell damage, turbulent shear stress conditions.	Sutera and Mehrjardi, (1975)
250	Test time 4 minutes. Release of haemoglobin indicating cell damage.	Sutera and Mehrjardi, (1975)
400/350	Threshold stress limit before damage	Sallam and Hwang, (1984)
600	Recalculated results of Sallam and Hwang, (1984) of threshold limit before damage.	Grigioni, Daniele et al., (1999)

Table 1.2, Estimates of the stress levels to damage red blood cells.

There may not be an exact figure for the start of red cell haemolysis and it is possible that haemolysis occurs at almost any stress but is either undetectable or is accommodated in the general metabolism of the body.

Figure 1.3 indicates a stress of 250 N/m² gives haemolysis of around 12% at an exposure time of 4 minutes. This is a short time as a blood pump may run for days, weeks or even years. The design of a pump to run for an

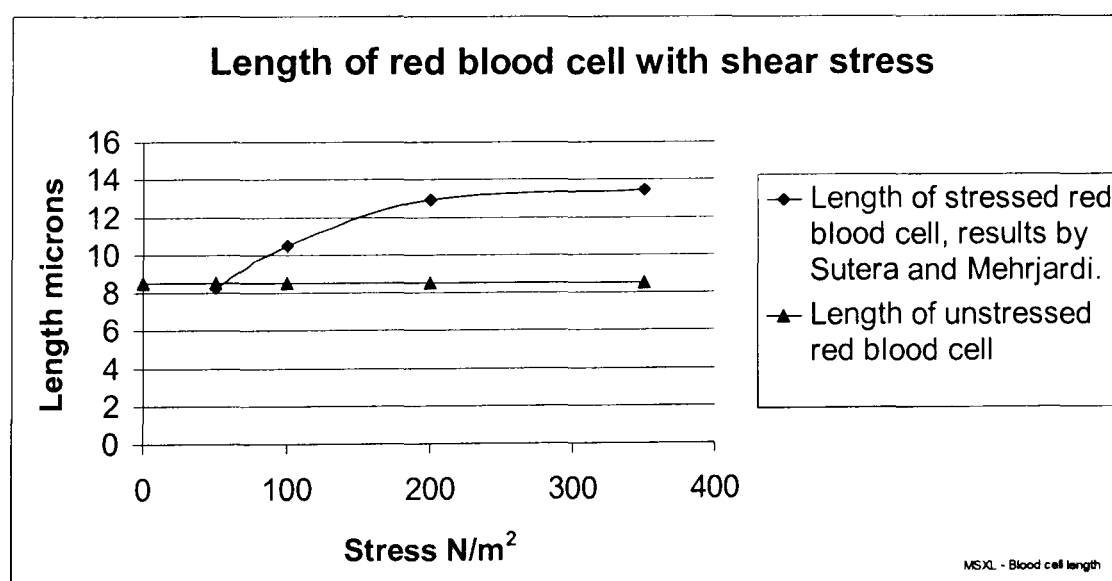
appropriate length of time will need shear stress levels lower than the figures quoted here.



Sutera S.P. and Mehrjardi M.H., 1975

Figure 1.3 Shear stress against percentage haemolysis.

From the above results it is difficult to determine a precise figure for the limiting stress for red blood cells and 250 N/m² will be used until a more accurate figure can be found. Results from Sutera and Mehrjardi, (1975), Figure 1.4, using a concentric cylinder viscometer and fixing the stressed red blood cells with a 1% glutaraldehyde solution, tend to confirm this as the rate of cell elongation against shear stress appears to approach a maximum around 250 – 300 N/m².



Sutera S.P. and Mehrjardi M.H., 1975

Figure 1.4, Red blood cell length Vs shear stress.

Sutera and Mehrjardi, (1975), conducted tests to determine the effect of stress on red blood cells.

The results, summarised in Table 1.3, indicate that when stressed up to 200N/m^2 for 4 minutes, the cells will regain their original shape when the stress is removed. Stress over this figure damages the cells with no shape recovery.

The length of time for each test is very short considering the time that a blood pump will need to run.

There is little or no agreement between the two results at 350 N/m^2 for 4 minutes. The comments indicate the cell condition is different for the same stress for the same time.

Stress N/m^2	Time (mins)	Condition of red blood cells	Comments
10	4	No signs of damage	Cells appear normal.
200	4	No signs of damage	Cells appear undamaged
200	4	Elongated whole cells	Cells are stretched, no apparent damage
350	4	Elongated with some fragments	As above but with greater stretching.
350	4	Fragments and crenulated cells	All cells showing severe damage.
450	4	Very elongated, many fragments	Cells damaged, many fragments
450	2	Elongated with some fragments	Cells damaged, with fragments and 'dumb bell' shaped cells

Table 1.3. Summary of cell stress tests, Sutera and Mehrjardi, (1975).

Leverett, Hellums et al., (1972), produced the results shown in Table 1.4. They indicate that the threshold level of damage differs depending on the

method chosen and it can range from '*relatively little*' to $4,000 \text{ N/m}^2$. (There is no precise definition here of what is meant by '*relatively little*'). The results indicate that for a 'long' time, (1000 seconds), and 'low' stress, ($60\text{-}25 \text{ N/m}^2$), '*relatively little*' haemolysis results per unit time. Increasing the level of stress to 150 N/m^2 , the threshold level of damage exposure time reduces to 100 seconds.

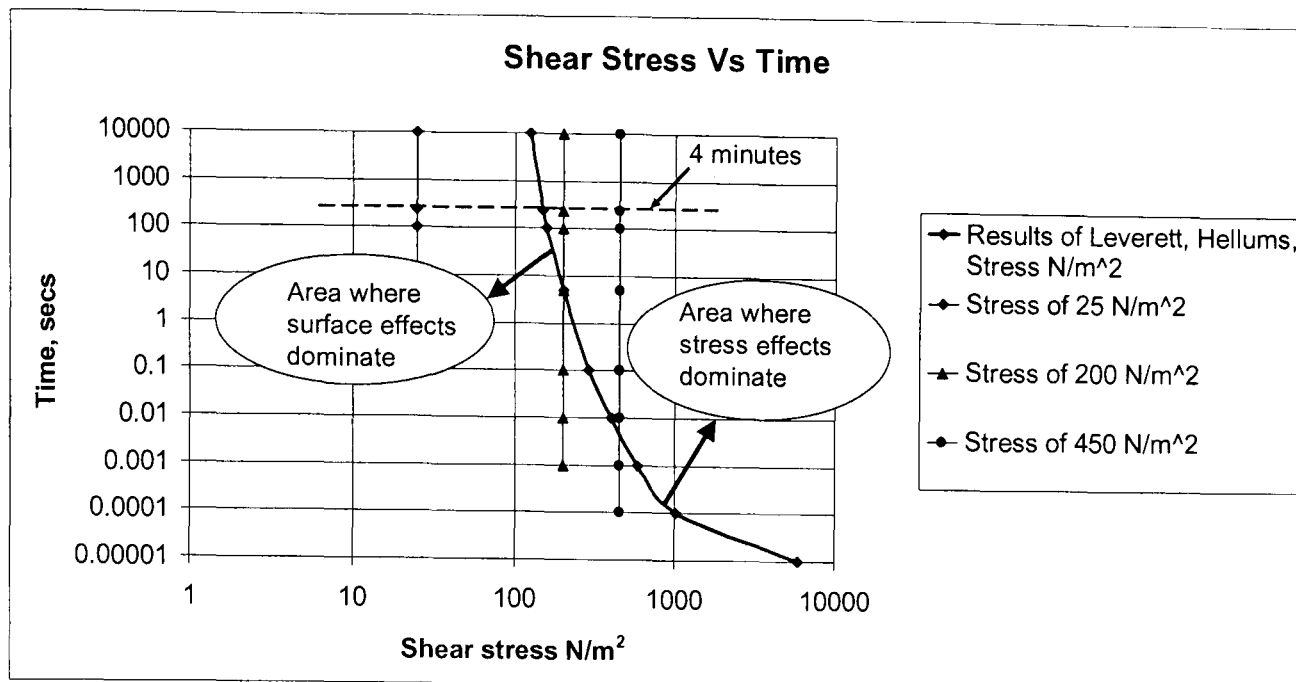
In terms of running a blood pump, the exposure times are extremely short, 1000 seconds is less than 17 minutes. The results with exposure times of 10^{-2} to 10^{-5} seconds are not considered applicable to a practical blood pump application.

Type of exposure	Order of magnitude of exposure time, sec	Threshold level of damage, N/m ²	Comments	References
Turbulent jet	10 ⁻⁵	4000		Forstrom (1969), Blackshear (1971)
Oscillating wire	10 ⁻⁴	560	Human and canine	Williams et al (1970)
Oscillating bubble	10 ⁻³	450	Human and canine	Rooney (1970)
Capillary flow	10 ⁻²	500	Bovine blood	Bacher and Williams (1970)
Capillary flow	10 ⁻²	450 - 700	Canine blood	Kashaviah (1970), Blackshear (1971)
Concentric cylinder	10 ²	150		Leverett and Hellums (1972)
Concentric cylinder max. stress 60 N/m ²	10 ² - 10 ³	Relatively little haemolysis per unit time	Surface effects dominate	Shapiro and Williams (1970)
Concentric cylinder max. stress 25 N/m ²	10 ³	Relatively little haemolysis per unit time	Surface effects dominate	Knapp and Yarborough (1969)
Concentric cylinder max stress 60 N/m ²	10 ³	Relatively little haemolysis per unit time	Surface effects dominate	Steinbach (1970) and Blackshear (1971)

Red blood damage by shear stress, Leverett, Hellums et al., Biophysical Journal, Volume 12, (1972), p 257 - 273

Table 1.4, A summary of shear stress and time on red blood cells.

Figure 1.5 shows a pictorial interpretation of the results of Leverett, Hellums, et al., (1972) from Table 1.4. The following observations are made from those results.



After Leverett, Hellums, et al., (1972).

Figure 1.5, The relationship of exposure time and the threshold shear stress.

The results of Leverett, Hellums, et al., (1972), Figure 1.5, can be divided into two parts. One part indicates that with 'low' shear stresses, surface effects are significant in determining the damage threshold. In this region, if surface effects can be minimised enough to be ineffective, the low blood stresses will enable a pump to run for an extended time period. The materials used in pump construction are therefore important.

In the other part of Figure 1.5 where the stress is 'high', stress effects are more important than surface effects in the potential for blood damage. Under these conditions a blood pump will not be safe to run for long time periods. Assuming a blood pump needs to run for years, stress levels and surface effects both need to be low to obtain a satisfactory pump performance.

From Figure 1.5, the limiting stress is time dependant. The stress limit will be 'low' or 'high' depending on how long the stress is sustained. Examples are shown where the stress lines for 200 N/m² and 450 N/m² run through both the 'stress dominant' and 'surface effects dominant' regions.

With a stress of 10 N/m² for 4 minutes, there is no apparent damage to the cells. This result is inside the 'surface effects dominate' region shown on Figure 1.5. This condition agrees with comments on Table 1.3.

With a stress of 25 N/m^2 , an apparent exposure time of greater than 10,000 seconds, (2.78 hours), is possible without stress effects being dominant.

If the trend line of Figure 1.5 continues asymptotically, the trend line may not cross the 100 N/m^2 stress line. Therefore a stress limit of 100 N/m^2 may be acceptable for $10^{8.5}$ seconds, (10 years). This tends to confirm the result on Figure 1.3 where there is no appreciable haemolysis at this stress level.

From Table 1.3, at a stress level of 200 N/m^2 for 4 minutes the cells are stretched. This result, shown on Figure 1.5, falls inside the 'stress effects dominant' region. For a stress of 450 N/m^2 for 4 minutes, cell damage is noted. This is not surprising, as from Figure 1.5, the stress effects are dominant down about 0.01 second.

From these results there may be a maximum blood stress of less than 100 N/m^2 for a satisfactory pump. However this figure is greater than the maximum blood shear stress shown in Appendix 2, 'Physiological Flow Parameters in the Human Circulation', which quotes a maximum stress of less than 6 N/m^2 in the arterioles.

The results from Figure 1.5 show that blood damage can be split into two parts, stress effects and surface effects. There will always be a stress effect in the surface effects region and there will always be surface effects in the stress region. The results of Leverett, Hellums, et al., (1972) should be interpreted as a limit for these characteristics but with low stress levels and suitable surfaces, significant blood damage should not occur.

The time that blood stays inside a particular pump type varies. Reul and Akdis, (2000), indicate residence times of 1 second and 1 ms for a centrifugal pump and axial flow pump respectively. Even at these times, these are not ideal pumps as they run at different stress levels, a centrifugal pump is quoted at $20 - 100 \text{ N/m}^2$ and a small axial pump is around 400 N/m^2 . It is assumed that pulsating pumps can have longer 'residence' times than this as with a 'beat' rate of 60 per minute, the shortest time will be 1 second and depending on the internal flow patterns, could be much greater. With a slower 'beat' rate, the residence times will be even longer.

1.3.5 White blood cells, (leukocytes).

White cells occupy less than 1% of the total blood volume and number about 5-9 thousand white cells per cubic millimetre of blood. The number of white cells is variable, increasing when infection is present and can reach high levels with severe conditions. A white blood cell has a life of about 13 days, about 10% of that of a red blood cell.

Under the microscope white blood cells appear as flat, circular cells but in practice can change shape constantly and have some intrinsic mobility. The shape changing ability allows them to pass from the bloodstream into organs and back again easily.

Leukocytes are important cells in that they protect against infection and proteins that are 'foreign' to the system. This protection can range from combating the infection due to a simple cold up to the rejection of a new organ. This protection process must be suppressed for a successful organ transplant to take place.

1.3.6 Platelets.

Platelets are smaller than a red blood cell and number about 250 to 300 thousand per cubic millimetre of blood. The lifespan of a platelet is about 10 to 12 days.

Platelets seal wounds by forming a plug at a wound site. They can be a problem if arterial blood flow is reduced when they build up on the inside of a damaged artery. (Berger, Hartwell et al., 1998).

Blood flowing through small or narrowed flow passages can influence the development of blockages. Normally this is a problem relative to age and lifestyle but with an artificial device implanted this can be a problem irrespective of age.

Bluestein, Niu et al, (1997), reported on the effects of blood flowing through a stenosis which is considered to be a narrowed, or partially blocked, section of artery. The stenosis produced turbulent flow at the throat with a recirculation region afterwards. Platelet deposition at the throat was low as the platelets were apparently swept away by the flow, which was up to five times greater than at the entrance to the throat. In the recirculation region, platelet

deposition was up to twice normal rate. In this area the shear stresses were low but with a high blood residence time creating favourable conditions for platelet deposition. Figure 1.6 below shows the general pattern of flow and platelet deposition.

Ramstack et al., (1979), indicated that platelet activation is a non-linear function where a large stress for a short time can produce significant platelet stimulation where the same stress for an additional length of time produces relatively little additional stimulation.

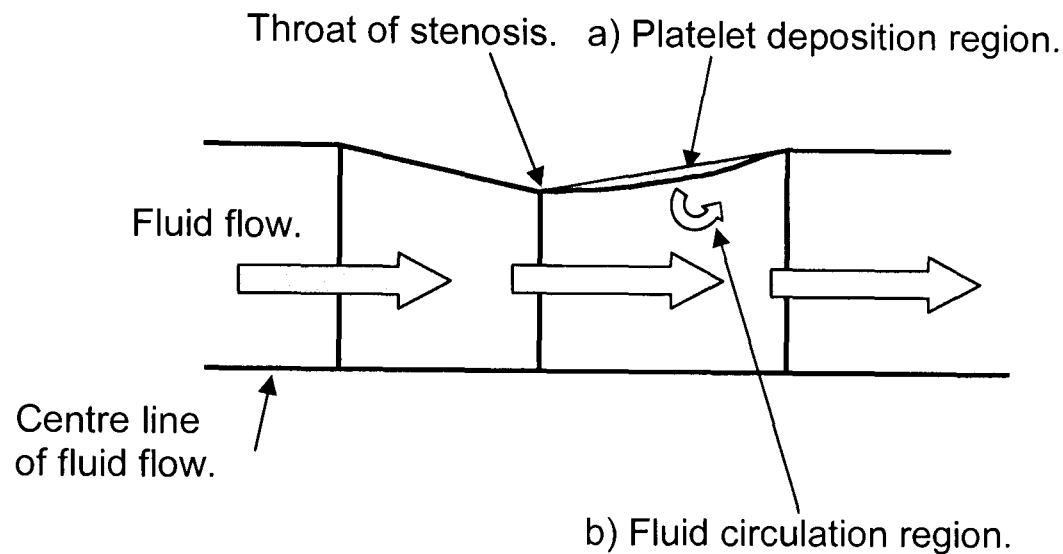


Figure 1.6 Half section through stenosis showing: -
a) Platelet deposition region and
b) Fluid circulation region.

Ramstack et al., (1979) and Wurzinger et al., (1983), indicate that platelets can be damaged by shear stresses of 10 to 50 N/m^2 . For stresses above about 15 N/m^2 , platelet death appears to start almost immediately but then decreases with time, indicating that some platelets are more prone to stress damage than others. Brown et al., (1975), suggested shear stress as low as 5 N/m^2 stimulated the platelet release reaction and platelet aggregation. Shear stress at 10 N/m^2 was reported to produce cell lysis which appeared to be linearly related to the duration of stress.

Brown and Leverett et al, (1975), indicated similar figures for platelet damage and lysis and demonstrated that platelets are sensitive to shear stress.

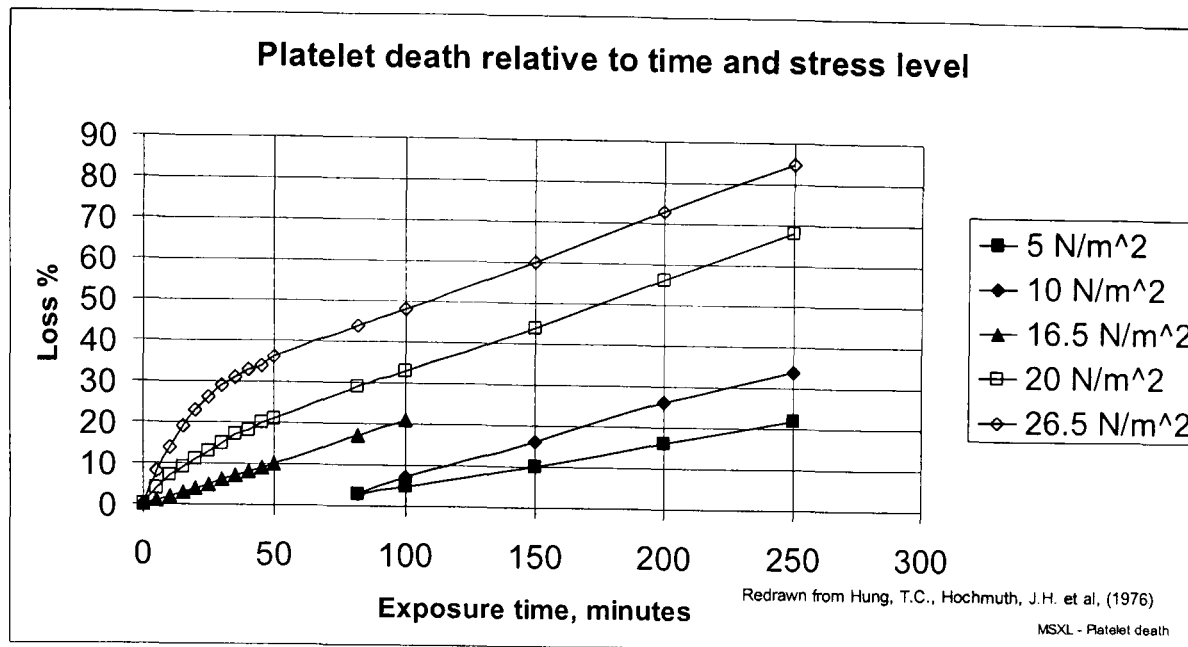
Analysis by Suter et al., (1988), showed that pulsed exposure resulted in more platelet aggregation than continuous exposure for the same total exposure time to stress. This indicated that both shear and velocity gradients might be responsible for enhancing platelet activation.

Hodges and Jensen, (2002), assumed that molecular forces attract a cell to a surface. With the cell drifting in the blood flow, the cell moves towards the surface. As the cell approaches the wall, the attraction forces increase and the cell contacts the wall. Depending on the force of the fluid flowing past the cell, the cell either stays fixed to and spreads along the wall, or moves along the wall by a rolling, or tank treading, motion or is peeled off by fluid flow past it.

Poiseuille, (1799 -1869), noted that during blood flow, cells moved towards the centre of a round tube if the flow was high enough. This may be why cells build up on a surface where there is little or no fluid flow and there is no cell adhesion in regions of fast fluid flow.

Hung et al, (1976), indicated that at stress levels of 5 and 10 N/m², there is no measurable cell death for the first 60 to 75 minutes. After this time a rising linear characteristic is shown for a time of up to 4 hours. Figure 1.7 shows these results. Higher levels of stress produced almost immediate results. These results may indicate that platelets become weaker during their life cycle and that the 'older' cells die first followed by the 'younger' cells later.

There were no results shown beyond 4 hours which is short compared to the time that a blood pump may be needed.



Redrawn from Hung, T.C., Hochmuth, J.H. et al, (1976).

Figure 1.7. Platelet death relative to time and stress level.

1.3.7 Significance of the above results.

The results shown in Table 1.3 are taken over a period of 4 minutes, or less, and during this time damage to red blood cells can be noted. A blood pump will operate for long periods of time. Any damage to blood cells caused by the pump will have to be at a level the patient can withstand.

The pump design must not produce fluid stress levels that are high enough to cause blood damage. From the above results the maximum fluid stress level to cause damage is approximately 200 – 250 N/m² for red cells and 10 N/m² for the platelets. Damaged cells will still be able to pass through the system and may block vital organs.

The results from Goldsmith and Turitto, (1986), (shown in Appendix 2), indicate that the maximum mean wall shear stress *in vivo* is less than 6 N/m². This is less than the stresses quoted above.

Mechanics of fluids theory shows turbulent flow produces stress in fluids. Assuming turbulent flow is indicated by a Reynolds number of 2000 or above, Goldsmith and Turitto, (1986), show turbulent flow in the ascending aorta at a Reynolds number of 3200 to 6000. The Reynolds number in the rest of arterial system indicates laminar flow

Changes in flow direction inside the pump must not produce flow patterns where blood is stagnant or in a continuous loop as reactions with

construction materials will be increased. The time that blood stays inside the pump must also be as short as possible.

1.4 Summary

The problem of pumping blood with mechanical pumps has been described indicating that shear stress can damage red and white blood cells. The length of time needed for change to occur can be as short as 4 minutes. If over stressed, the cells do not recover and the clotting and aggregation of blood cells is automatic once it is initiated.

The maximum stress *in vivo*, of about 6 N/m^2 , is very low compared to the stresses that can be produced by pumping blood. Red cells may survive a stress of 100 N/m^2 for up to 10 years but further work needs to confirm this figure. White cells appear to have a lower damage threshold at 5 to 10 N/m^2 before aggregating. A blood pump will have to produce stress levels less than 10 N/m^2 for a satisfactory life following these results.

A blood pump should ideally produce laminar fluid flow to reduce blood stress levels and cell damage. There should be no recirculation flow or flow stagnation of the blood inside the pump. One of the design criteria must be to ensure that the time the blood stays inside the pump must be as short as possible to reduce surface reactions with the blood.

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Zaroff, J. et al., Consensus Conference Report. Maximizing Use of Organs Recovered From the Cadaver Donor: Cardiac Recommendations.
March 28-29, 2001, Crystal City, Va. *Circulation*, (2002); 106:836-841.

Chapter 2. Review of heart pumps.

2.1 Introduction.

To repair or replace a faulty human heart has long been the dream of the medical profession. As the body must have a blood supply for survival, stopping or removing the heart for repair has not been an option if the supply of blood to the body is stopped. Pumps have now been made which take over all, or part, of the pumping function of the heart enabling repairs to be carried out on the heart without the patient dying. These pumps range from pulsating pressure pumps, that imitate the heart, to constant pressure pumps. Not all pumps have been successful, fluid dynamics problems have damaged blood by excessive shear stresses and unsuitable flow patterns have allowed thrombi to form. The basic pump types are described in Appendix 3.

Most blood pumps are intended for adults but the low flow rates required for a child pose an increased thrombus risk when using adult size devices due to the reduced blood flow.

The Table 2.1 shows a brief history of ventricular assist devices.

1812	Legallois – first concept
1849	Loebell – isolated kidney model
1854	Claude and Bernard – <i>foie lève</i> experiment
1848 - 1858	Brown and Sequard – demonstrated the need to oxygenate blood
1868	Ludwig and Schmidt – achieved extracorporeal oxygenation
1882	Von Schroeder – first steady bubble oxygenator
1885	Frey and Gruber – first film oxygenator
1916	McLean discovered heparin
1928	Dale and Schuster – first diaphragm pump
1934	DeBakey – introduced a simple roller pump
1937	Gibbon – described a heart lung machine
1954	Gibbon – first human open heart operation
1957	Akutsu and Kolff – implanted two pumps in a dog
1961	Dennis – performed left heart bypass
1962	Kolff and Mouloupoulos – introduced the intra aortic balloon pump
1963	Spencer – supported a girl after a septal defect
1964	Loitta – first implantation of a pulsatile LVAD
1966	Kantrowitz – first clinical application
1968	Raffert – reported on a centrifugal blood pump
1969	Cooley – implanted the first artificial heart
1973	Soeter – described the use of extracorporeal life support
1974	Bartlett – use of extracorporeal membrane oxygenator
1980	Pollock – first use of intra aortic balloon pump in children
1980	Thoratec ventricular devices
1982	DeVries – implantation of first Jarvik 7 artificial heart
1983	Veasy – use of small balloon catheters
1984	Novacor LVAD used as a bridge to transplant
1987	First use of a centrifugal Biomedicus pump
1988	Berlin heart becomes available
1989	Extracorporeal Life Support Organisation founded
1992	First Heartmate pumps implanted
1992	Berlin heart offered for the first time
1993	First wearable control system
1994	Medos VAD in clinical use
1994	Abiomed BVS 5000 VAD becomes available
1994	Medos VAD first implantation

Table 2.1. Brief history of ventricular assist devices

<http://www.health.gov.mt/impacedcard/issue9/2454/2454.htm>

Fuchs A, Netz Z., (2002)

2.1.1 The generations of heart pumps.

The first implantable ventricular assist devices were produced in the 1960's and it has taken 15 – 20 years to reach true clinical applications. These pumps were pulsatile devices imitating the action of the human heart. They were the first generation of artificial hearts or blood pumps and many were total heart replacement pumps where the native heart had been removed. These pumps had often had problems with size, patient acceptability, power sources, and blood clot generating properties.

The second generation pumps include constant flow pumps. These run in parallel with the heart which is still operating and were intended to solve some of the problems of the first generation as they were smaller, lasted longer and gave fewer problems. Power and control systems are starting to become more acceptable with the possibility that the patient could be 'unfettered' from the confines of the hospital and the wall power socket. Some power packs allow the patient to return home and resume light work, producing a general improvement in patient condition.

The third generation pumps, described by Bourqu and Gernes et al, (2002), have yet to come into general use and are intended to use fully implanted pumps and control systems. Some have non contact bearings using a magnetic or fluid levitation system and the service life is expected to be at least ten years as outlined by Takatani S., (2002).

There have been many types of blood pumps proposed, or made, and some of these are shown in Appendix 3.

Table 2.2 shows the generations of left ventricular assist systems, (LVAS), and general characteristics. Reul and Akdis, (2000) discuss a range of blood pumps with advantages and disadvantages, quality of life, reliability and cost effectiveness.

LVAS		Advantages	Disadvantages
First Generation	Pulsatile pumps. Pumps are air or electrically driven Bulky internal unit. Cumbersome external equipment.	Capable of generating flows that are close to the natural flow characteristics of blood.	Poor durability. Limited mechanical life. Risk of thrombotic events. Large implanted size means only a proportion of patients is suitable. May require a substantial amount of anti-coagulant drugs.
Second Generation	Non-pulsatile pumps. Use of contact bearings and the rotational force of an impeller to maintain blood flow. Two major types: centrifugal or axial flow pumps.	Smaller than first generation devices, hence a wider range of patients can be treated. Lower risk of infection than first generation devices.	Bearings generate heat, increasing the risk of thrombotic events. Higher risk of red blood cell destruction. Limited mechanical life. May require amounts of anti-coagulant drugs.
Third generation	Non-pulsatile pumps No contact bearings Two major types: magnetic ¹ and hydrodynamic ² levitation.	Lack of contact bearings reduces the risk of thrombotic events. Lack of wear enhances durability Less red blood cell damage.	Magnetically suspended systems may be more complicated and therefore more prone to failure.

Emerging Growth Capital Pty Ltd., (2004)

¹ Magnetic levitation involves the suspension of the impeller by magnetic forces.

² Hydrodynamic levitation involves the suspension of the impeller by fluid forces.

Table 2.2 The classification of blood pumps.

The pumping principles of first generation pumps are listed in Table 2.3 together with the power source. This shows a connection between some pumps and power source and that many types of power source have been attempted. Many do not appear to have reached application testing.

<u>Pumping principle.</u>	<u>Energy conversion.</u>
Displacement pumps	Electro – mechanical.
Hydrodynamic pumps.	Electro – magnetic.
Combinations.	Electro – hydraulic.
	- indirect.
	- direct.
	Thermo – mechanical
	Radio – isotope
	Biological – mechanical.

Hetzer, R., Hennig, E., Loebe, M. (editors), (1997)

Table 2.3 Types of ventricular assist devices.

2.1.2 Explanation of pump types.

Displacement pump.

This pump gives a pulsating blood flow similar to the biological heart. A moving diaphragm, with valves controlling the direction of blood flow, achieves the pumping action.

Hydrodynamic pump.

This gives a constant pressure blood flow. The flow is produced by a rotating impeller and there are no valves to control blood flow direction.

Electro – hydraulic pump

This is a pulsating pump with the diaphragm driven by hydraulic means. The non blood side of the diaphragm is acted upon by hydraulic pressure generated external to the blood pump.

Radio- isotope pumps

Public concern and cost for a radio - isotope pump have eliminated this as an implantable power source at present.

Biological – mechanical pumps

This uses skeletal muscle to drive a plunger arrangement to drive a diaphragm, which then pumps the blood. The muscle is activated by an electronic device similar to a cardiac pacemaker. Skeletal muscle is not the same as cardiac muscle but Salmons and Jarvis, (1990), et al., found that

skeletal muscle could be transformed into something approaching cardiac muscle by applying a low frequency electrical stimulation.

Figure 2.1 shows a typical arrangement of a muscle powered pump.

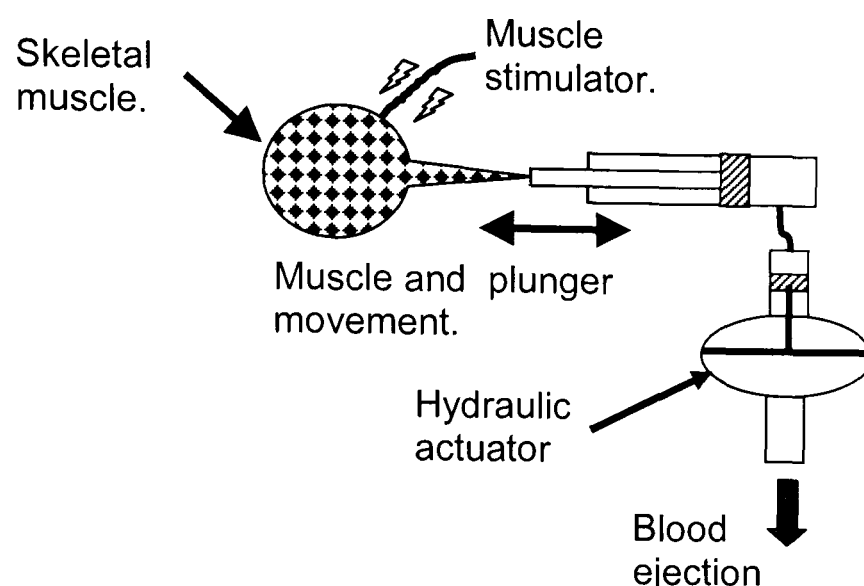


Figure 2.1. Arrangement of a muscle powered blood pump.

Pumps available or in development are shown in Appendix 3. The characteristic differences between the two main pump types are listed in Table 2.4.

Blood pump	Pulsatile	Non-pulsatile
Size	Large	Smaller
Mechanism	Complex	Simpler
Control	Complex	Simpler
Valves	Two	None
Compliance chamber	One	None
Implantability	Complex	Simpler
Overall system	Same	Same
Cost	Higher	Lower
Physiological acceptance	OK	NA

Takatani S, (2002)

Table 2.4. Comparative features of pulsatile and non-pulsatile blood pumps.

Technical problems of these devices, listed in Table 2.5, are covered in detail elsewhere.

<u>Durability</u>	<u>Characteristic</u>
Membrane	Flex life, degradation.
Gearing	Wear, corrosion.
Bearing	Wear, seizure.
Sealing	Wear, seizure.
Encapsulation	Fluid permeation.
<u>Biocompatibility</u>	
Blood contact	Thrombus formation, cellular trauma, trauma.
Tissue contact	Cell ingrowth, infection.
Body fluids	Chemical degradation.
<u>Power supply.</u>	
Batteries / TETS	Heat dissipation, efficiency, cycle life.
<u>Control</u>	
Electronics	Reliability
Sensors	Long term stability
<u>Size and weight</u>	
Compactness of design.	Device volume / pump output.
Anatomical fit	Location of components
<u>Fail safe features</u>	
	No fail safe features noted

Hetzer, R., Hennig, E., Loebe, M. (editors), (1997).

Table 2.5. Technical problems with ventricular assist devices.

2.1.3 Physiological adaptation

The human heart is a pulsatile device and the arteries leading from it are elastic. Each heart beat causes the arteries to expand and then contract as the pressure pulse passes.

Muria and Ichikawa, et al., (2003) report that the Baylor College of Medicine, Houston, Texas conducted tests on 5 calves to determine if the change from pulsatile to non-pulsatile blood flow had any serious effects.

Haemodynamic data was established before each calf was implanted with a non-pulsatile biventricular assist system and the analysis was repeated at intervals. Initially the results did not correlate but the final results indicated that the calves had adapted to the new conditions. If this procedure gives the same results in humans, there should be no adaptation problems with continuous flow pumps compared to pulsatile flow pumps.

2.2 Pulsatile flow pumps (displacement) types.

This section describes the type of pump that delivers fluid in a series of pulses, similar to the human heart. The amount of fluid delivered will depend on the size of the pump and the pulses / min.

2.2.1 Peristaltic or roller pumps.

Method of operation.

This pump has multiple rollers pressing against flexible wall tubing. Figure 2.2 shows the arrangement of the rollers and tubing. The rollers move relative to the tubing which is arranged in a near circular arrangement to enable maximum roller contact. These are similar to the pumps manufactured by Watson-Marlow Bredel Pumps. The rollers separate the fluid into 'packets' that are delivered to the user. This type of pump is used in heart-lung machines and kidney dialysis procedures.

The 'finger' pump is similar where a series of fingers press down sequentially on to a straight length of tubing creating the 'packet' of blood delivery arrangement.

The output of this pump will vary depending on the rotational speed of the roller assembly and the tubing diameter.

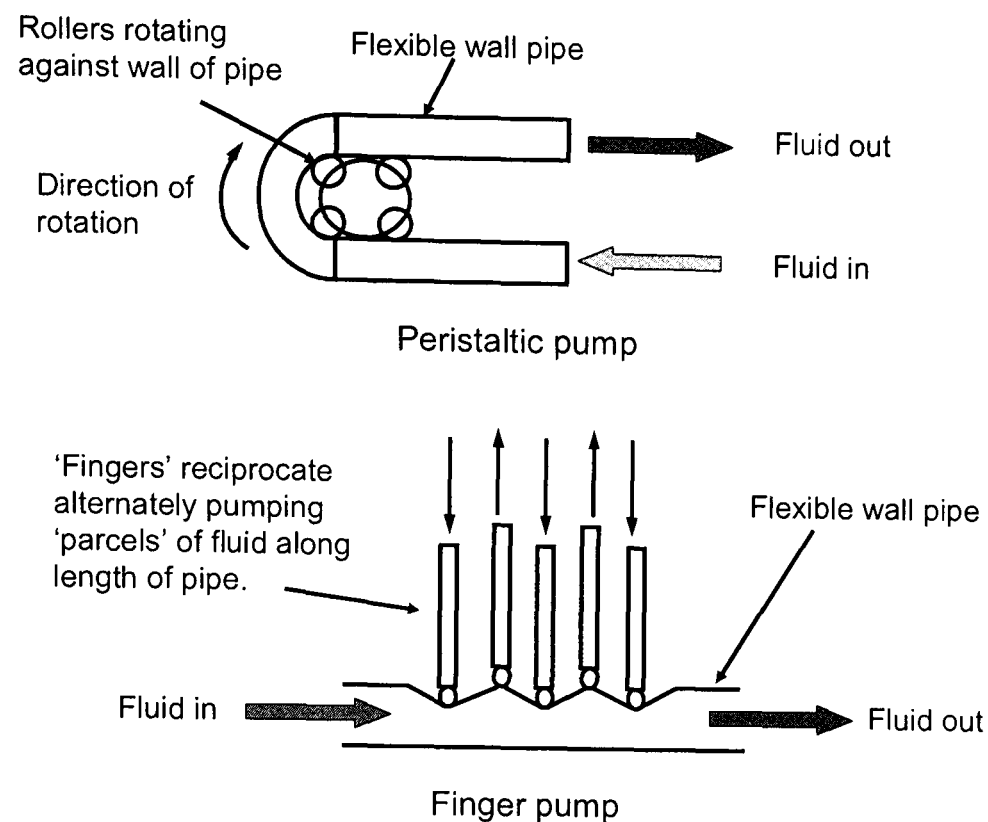


Figure 2.2. Peristaltic and finger pumps.

Advantages and disadvantages of peristaltic or roller pumps

Advantages.

Low cost of tubing and reliability. They are simple to use, cheap, easy to set up and replace. The amount of blood delivered by the pump can be changed by changing the rotation speed of the rollers for a given size of tubing.

Disadvantages.

Some blood damage and particulate spallation, (small particles breaking away from inside the tubing).

'Old' pumps may suffer from reduced output, as delivery is dependent on the tubing expanding fully after each roller has passed. The pump must be replaced before the tubing cracks otherwise blood could leak out or air could be drawn into the system. Both these will be detrimental to the patient.

Pump use is limited to about 4 hours due to blood damage and the need for high doses of anticoagulant.

The blood is passed to and from the patient using needles inserted into veins or arteries which has the risk of infection.

This method is confined to hospital use as staff and specialised equipment are needed for satisfactory operation.

2.2.2 Intra-Aortic Balloon Pump.

This is not a heart replacement device but it is used as a short-term assist device. It is removed as soon as possible.

Method of operation.

A balloon placed inside the aorta, downstream of the coronary arteries, is inflated at the end of left ventricle contraction. The blood flow through the aorta is blocked and a pulse of 'high' pressure blood flows through the coronary arteries. The balloon then deflates allowing normal blood flow to the body. Hetzer, R., Hennig, E. et al., (1997) explain the concepts behind this method of blood pumping

This is a gas powered, (helium), temporary device. Careful control of the timing of inflation and deflation is necessary and it is for hospital use only. The pump is used for a short a time, (hours), and is removed as soon as the patient has recovered sufficiently.

2.2.3 Bi-ventricular blood pump, (Abiomed 5000 blood pump.)

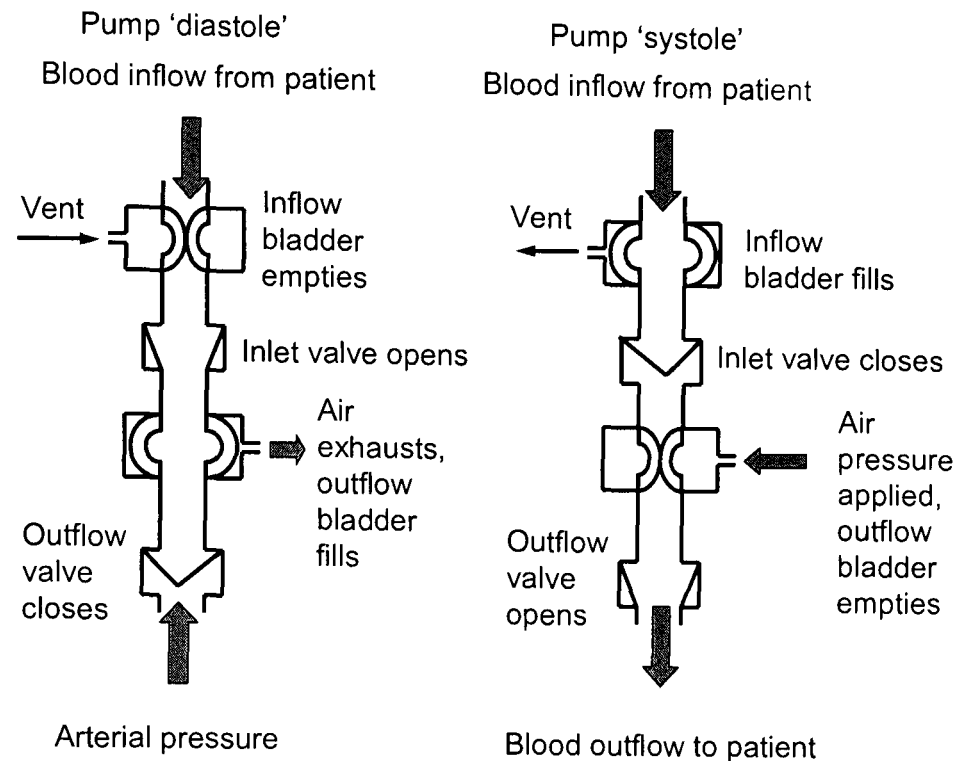
This is a heart assisting device following surgery and it is not an implantable device.

Method of operation.

The pump consists of a chamber containing a bladder, made from smooth surfaced polyurethane, which passively fills with blood from the patient. The pneumatic system externally pressurises the bladder and pumps the blood to the patient with valves controlling blood flow direction. The bladder volume is 100 ml. and the pump maximum flow rate is about 6 l/min.

The pump stroke volume is maintained at about 80 ml and computer control allows full automation of the system. There is usually no need for staff intervention during normal operations. Flow through the pump can be reduced to a minimum of 0.5 l/min as the patient improves. (Wassenburg, 2000).

This pump is intended for short-term use, about 14 days, to allow recovery from surgery and can be used singularly on either left or right sides of the heart, or two pumps to assist both sides of the heart simultaneously. Figure 2.3 illustrates the operation of this pump.



Abiomed Inc., Danvers, MA, USA.

Figure 2.3. Abiomed BVS 5000 blood pump.

The pump cannot be used outside the hospital environment and there is a risk of infection arising from the pipes passing through the patient's skin. (Wassenburg, 2000).

2.2.4 Other pulsatile pump types, general description.

These are diaphragm pumps driven by air, hydraulic, mechanical or electrical methods. Valves are used to control the blood flow direction but these can be a problem as they can promote blood clots endangering the patient.

Mussivand T. et al., (1996), indicate that careful design and assembly of the diaphragm is necessary and improvements in prediction techniques need to be used to avoid problems

With all pulsating pumps, sufficient 'wash out' at the junction between the diaphragm and the pump housing is necessary as with the increased length of time that blood stays inside the pump, the greater chance of surface reactions. Figure 2.4 shows this problem.

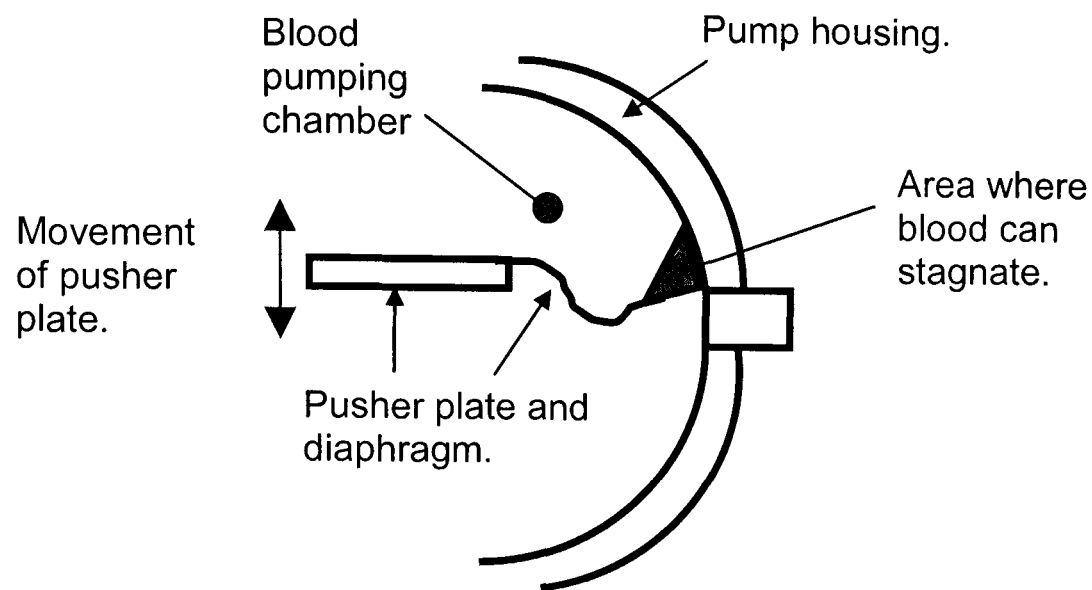


Figure 2.4 Showing the possibility of blood trapped in a poor 'wash out' region of a pulsating type pump.

Fuchs and Netz, (2002), describe the particular problems related to heart assistance with children. Many of the existing pulsatile pumps have pumping chambers that are too large for a child. Correct size is important as a pump that is too large can result in a poor wash out at the pumping surfaces leading to the development of thrombi with consequent problems.

2.2.5 Heartmate pump.

One of the original positive displacement pumps was the Heartmate described by Gagnon et al, (2000). Two versions were produced, a vented electric, VE-LVAD, and pneumatic, IP-LVAD.

Both used a two part titanium housing with entry and exit ports. The internal surfaces of the housing were textured by sintering titanium spheres to the surface to present a compatible surface to the rest of the blood as it flowed through. Each port contains a porcine valve controlling the direction of

blood flow. The pump uses a textured surface polyurethane diaphragm to attract blood cells. Using these methods a uniform tissue lining is formed through the entire internal surface of the pump.

The pump can operate in a fixed-rate or automatic mode. In addition, the IP-LVAD version can be operated in a synchronous mode triggered by the patients own electrocardiographic signal.

In the automatic mode, power is required only to pump the blood out of the pump. The pump fills passively, where blood flows into the pump from the patient. A position sensor detects when the pump is 90% full and power is applied to push the blood out.

Each pump has a maximum stroke of 83 mm with a maximum pumping rate of 140 beats / minute for the IP-LVAD and 120 beats / minute for the VE-LVAD. This gives flow rates of 11.6 and 9.96 L/min respectively. The pump weighs about 680 grams which is about twice the weight of the human heart.

There is only one size of the Heartmate pump and due to this, it is not possible to implant into infants / children or adults who are of small stature.

2.2.6 Power supplies.

The HeartMate pump requires careful installation in terms of power as the patient is effectively 'fixed' to a power supply, with power cables or pneumatic pipes passing through the skin with the chance of infection at the entry site.

Developments have allowed some HeartMate patients to leave hospital after careful training in the use of portable systems. Of those patients allowed to leave hospital with implanted pumps running, Korfer et al, (1997), report that some were strong enough to return to light work. With the VE-LVAD, the power supply consists of batteries carried on the belt and a control system in a shoulder holster. With this application there are two percutaneous leads coming out of the patient, one is the electrical power cable and the other is the vent pipe to equalise the air pressure under the diaphragm as it moves inside the pump.

With the IP-LVAD, the patient is connected to an external drive console, weighing around 33 kg, installed in a portable cart which limits patient mobility. This pump has only one pipe for the pneumatic supply. (Myers, Macris, 1995).

One of the later portable power systems, the TETS system, (Transmission of Energy Through the Skin), is described later in Chapter 2.4.2.

2.2.7 Volume compensation methods.

Pusher plate pumps always have a varying volume under the pusher plate as the pump operates as a volume is created and destroyed on the non-blood side of the diaphragm

With electrically driven pumps, as the diaphragm moves, the pump will not operate properly if the volume under the diaphragm is not vented, as the pump will be trying to compress / expand the air / gas trapped underneath. This will affect the efficiency of the pump and cause increased loading on the operating system. Pneumatically driven pumps do not have this problem as the volume under the plate is pressurised to pump the blood, and evacuated to allow blood to enter the pump

With diaphragm pumps, the problem of venting the volume under the diaphragm plate can be solved by:-

- a) Allow a vent pipe to exhaust to the atmosphere. This means a pipe passing through the skin of the patient with the additional risks of infection.

- b) Fit the pump with a variable volume compliance chamber or balloon. These are fitted internally to the patient and are filled with an inert gas. If they leak, they have to be refilled which can mean a surgical operation. There is the problem of where does the gas go to after it has leaked out? This method of volume compensation is rarely used today.

Figure 2.5a shows the operation of a pneumatically operated blood pump and Figure 2.5b shows a volume compensation vent with an electrically driven pump.

Figure 2.5. HeartMate pump.

Mode of operation of a positive displacement pump giving pulsating blood flow.

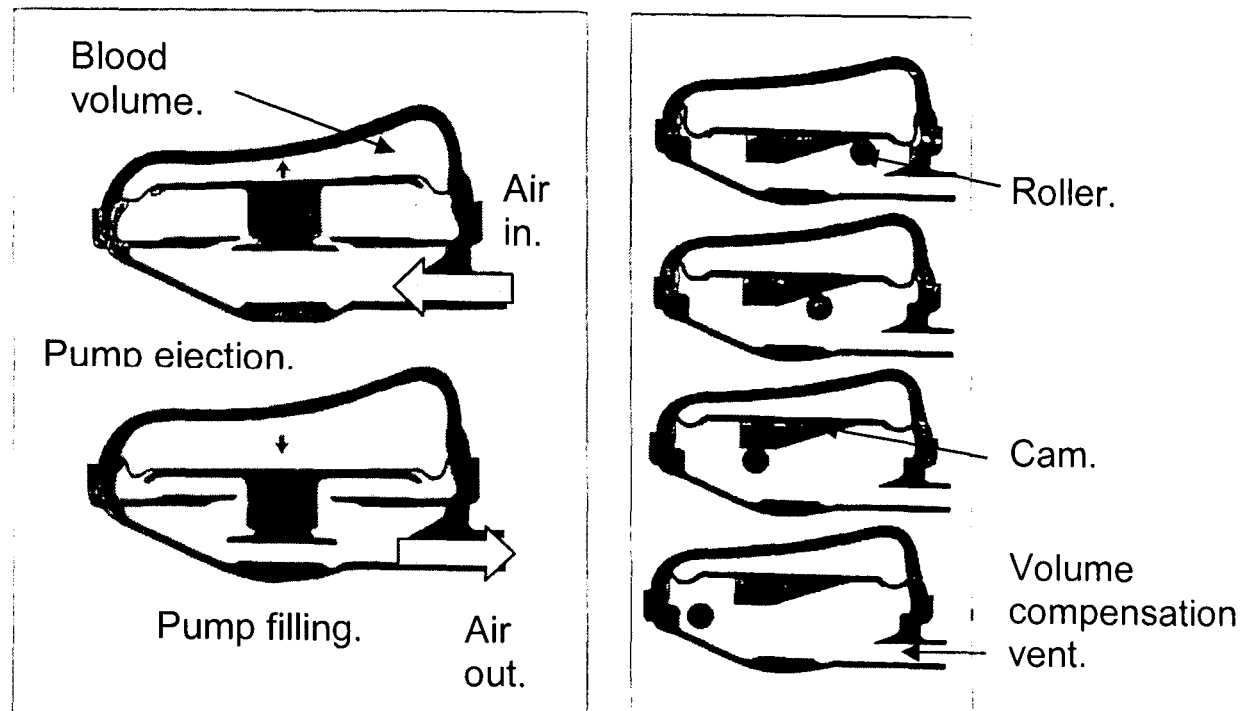


Figure 2.5a.
Pneumatic version.
Alternate air pressure application,
and release, drives diaphragm up
and down to pump blood

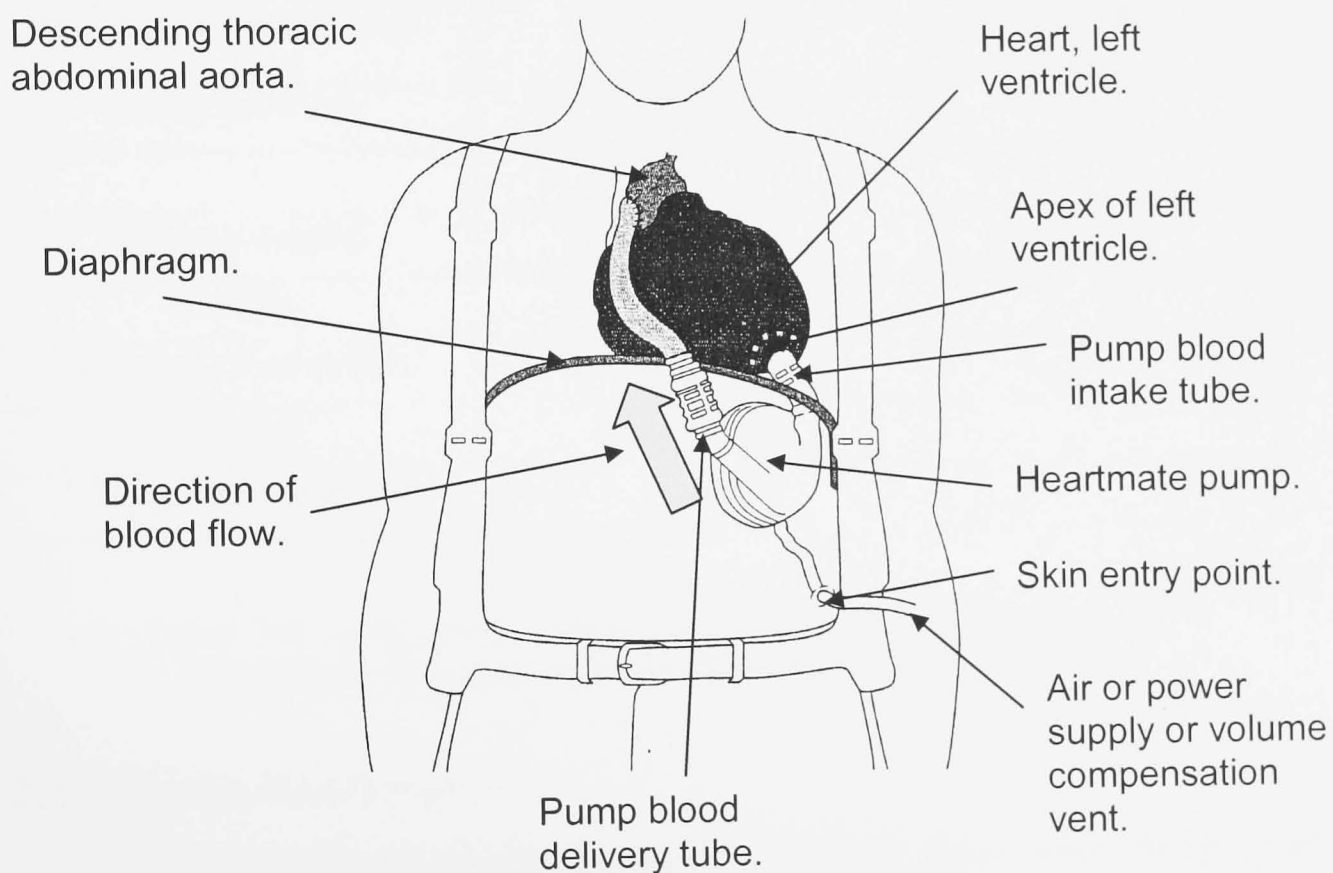
Figure 2.5b.
Electric motor version.
Roller contact with
circular face cam allows
diaphragm to rise and
fall and to pump blood.

2.2.8. Installation of a blood pump.

The original approach to help patients with heart problems was to remove the biological heart completely and replace it with an artificial heart if a transplant was not available. An alternative approach is to retain the biological heart and to fit a blood pump to act as an assist or bypass device. This can be used in the short term to allow the patient to recover from surgery or to allow a weakened biological heart to recover or to keep the patient alive long enough to receive a heart transplant. Many of the pumps produced today use this method. Figure 2.6 shows this method of installation using the Heartmate pump. This pump, like many others, operates in parallel with the biological heart by having the pump implanted between the apex of the left ventricle and the descending thoracic abdominal aorta. The pump receives low-pressure blood from the left ventricle and ejects it into the high-pressure

systemic circulation. Frazier, (1995), reports that the longest support period by one person on of these pumps was 503 days.

There are no published details of how the pump is installed inside the patient and held in place. The body has a tight fitting skin and the internal organs fit into place around each other. The installation of a grapefruit sized piece of machinery with the pipes or cables must cause problems with displacement. There must also be methods of fixing the pump securely in place to ensure that there is no strain on the pipes, ventricle and aorta and movement relative to other organs that could cause problems.



Loebe et al, (2000).

Figure 2.6. HeartMate pump. Installation inside the chest.

Table 2.6 shows the results of a multi-centre evaluation of the vented HeartMate pump. These show that one patient lived for nearly 700 days with the pump implanted and that 84% of patients were still alive 1 year after implantation. Some adverse events are shown, infection at 40% is the highest cause of problems. Problems that may be traced directly to the pump are low at 5%, neurologic dysfunction, (which is interpreted as a stroke), and 6%,

thromboembolism, (which is interpreted as blood clots blocking a blood vessel).

	LVAS supported patients.	Control, (not supported With LVAS)
No. of patients (M, F)	280 (232, 48)	48 (40, 8)
Age (median)	11-72 (55)	21-67 (50)
Pump duration, days (mean)	1-691 (112), 54/280 (19%) >180.	
Survived	198/280 (71%)	
Transplanted	188/280 (67%)	16/48 (33%)
Post-transplanted 1 year survival	158/188 (84%)	10/16 (63%)
Device removed	10/280 (4%)	
Adverse events		
Bleeding	31 (11%)	
Neurologic dysfunction	14 (5%)	
Infection	113 (40%)	
Thromboembolism	17 (6%)	

Takatani S, (2002)

Table 2.6, Multi centre clinical evaluation of the vented electric HeartMate Thoratec pump as a bridge to transplantation.

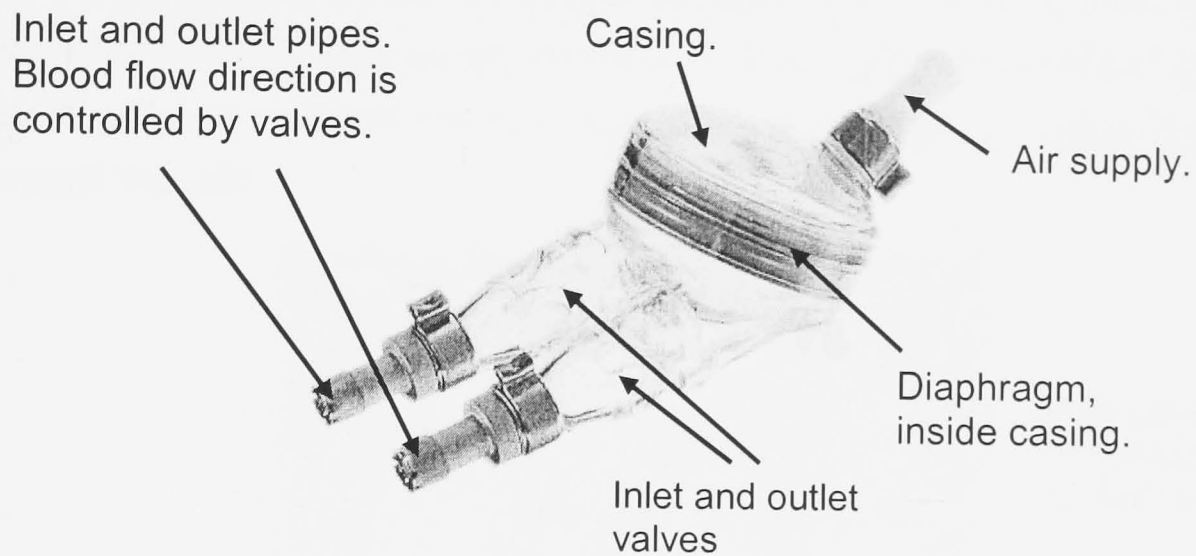
2.2.9 Berlin Heart Pump.

This is a pneumatically powered diaphragm pump with valves. The bloodlines run through skin entry sites and the pump can be supplied in a range of sizes suitable for a small child up to an adult. It is intended for short-term use only as a bridge to recovery or for a patient awaiting transplant.

For a child to require heart assistance, the device has to be sized accordingly. This was emphasised with the cases of a boy in 1998 and a girl in 2000 when both were fitted with small pumping systems provided by Berlin heart models. Both were kept alive until transplants could be found.

The Berlin heart pump, first offered in 1992, is a pump for children. It consists of a polyurethane housing with integrated diaphragms forming a

continuous blood-contacting interior. The three layer drive diaphragm is connected to a circular plate to form the air chamber. The multi-layer construction of the diaphragm membrane gives a diaphragm that is relatively thin but with high flexibility at the diaphragm-housing junction. Figure 2. 7 shows the general configuration of the pump.



Fuchs A., Netz H., (2002)

Figure 2.7. The general arrangement of the Berlin heart pump.

The internal blood contacting surfaces are smooth with inner surfaces coated with heparin. The blood chambers and outlet ports are transparent so that blood flow and any thrombus build up inside the pump can be noted. Two basic models are produced with various pumping capacities. They are listed in Table 2.7.

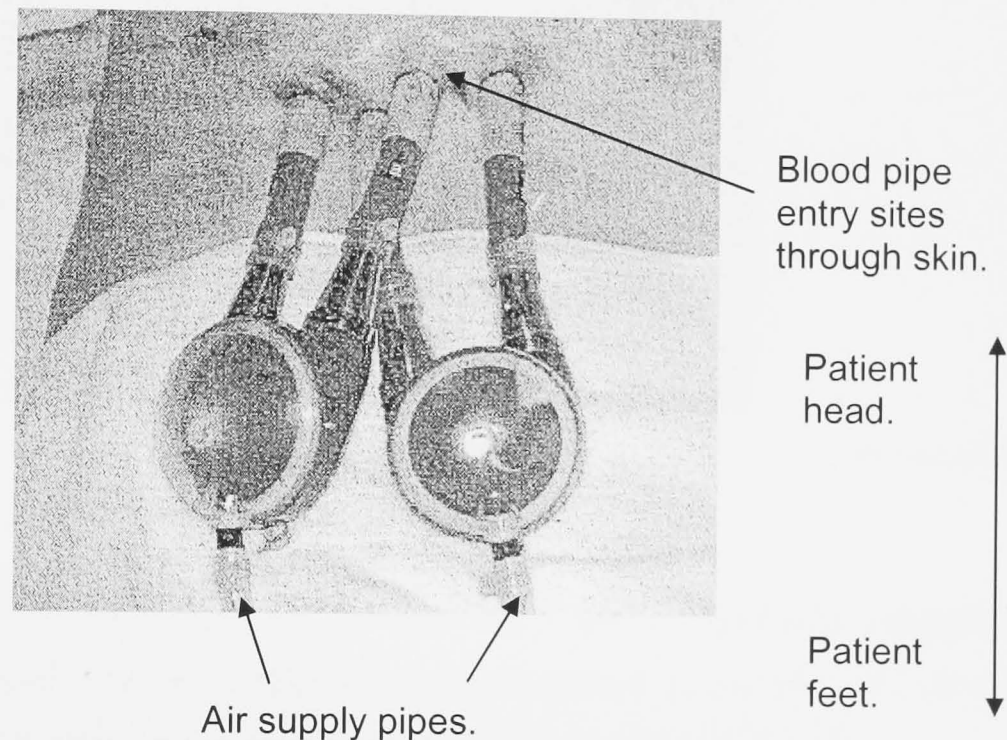
<u>Pump size.</u>	<u>Stroke volume ml.</u>	<u>Valve diameter, mm.</u>	<u>Valve type.</u>
Adult	50 and 60.	12	Bjork Shiley tilting type.
Child	12, 15, 25 and 30	12	Polyurethane tri-leaflet

Table 2.7. Details of Berlin heart types.

Figure 2.8 shows the installation of two Berlin heart pumps. Blood exits and enters the pump through pipes passing through the skin.

This pump can be used outside the hospital environment after suitable training and portable drive units have been developed to enable the patients to live relatively normal lives outside the hospital. The units all feature rechargeable batteries, together with computer controls and multiple redundant systems to ensure patient safety.

If the patient is outside hospital, frequent checks must be made where the pipes enter the body as any infection must be dealt with immediately. The infection risk is a problem with all systems that use pipes or cables passing through the skin.



Perfusion 2000; 15: pp 387-396.
Drews.T, Loebe, M et al, (2000)

Figure 2.8. Installation of two Berlin Heart pumps.

It should be noted that the power requirements for children are higher than those for adults as, due to the small size of the pipes used for children, the resistance to fluid flow is high during the operation of the pump and positive pressures of up to 350 mmHg and negative pressures of 100 mmHg at pumping rates of 140 beats/min are used. The drive unit has to be capable of giving this performance satisfactorily. (Drews and Loebe, et al, 2000).

2.2.10 AbioCor Heart pump.

This is an implanted electric pump and is designed to replace the biological heart and give patients a new, limited, lease of life. Figure 2.9 shows the external view of the pump.

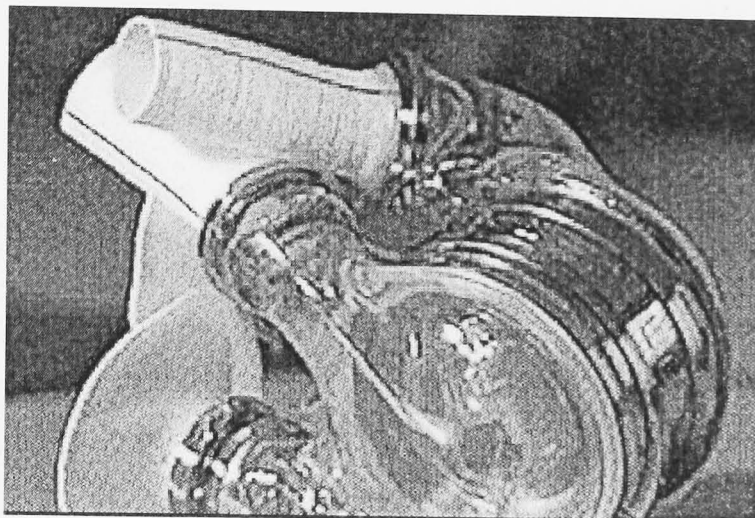


Figure 2.9. AbioCor pump - external view.

Abiomed, (2002)

The pump is a total artificial heart, (TAH), where the patient's ventricles have been removed and the artificial heart is attached to the remains of the atria. The patient is completely dependant on the pump for his/her life.

The pump is a self contained replacement TAH and is an electrically powered hydraulic pump powered by an internal battery which in turn is charged through the skin by an external battery pack. An internal electrical coil implanted just under the skin collects power from an external coil positioned above the internal coil. There are no holes through the skin therefore no chance of infection with this method. This is the TETS method that will be described later.

The pump operates by hydraulic pressure generated by a constant speed electric motor driving a rotary impellor and the hydraulic fluid is switched from left to right ventricle and back again by a rotary valve driven by a separate electric motor. Each ventricle is externally compressed by the hydraulic fluid to pump blood around the body as required. This is different to

the human heart where both ventricles contract at the same time. An implanted controller monitors and controls the speed of the artificial heart.

Figure 2.10 shows the internal hydraulic circuits of the pump. (http://www.chfpatients.com/implants/artificial_hearts.htm)

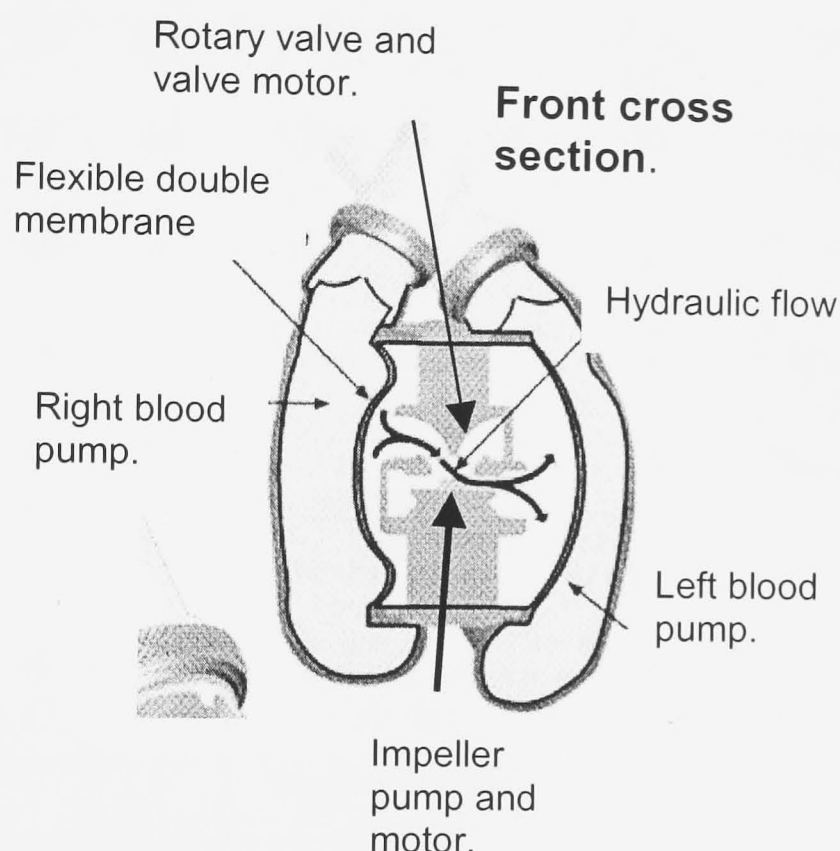


Figure 2.10. AbioCor pump hydraulic circuits and rotary valve.

Abiomed, (2002)

The pump has a mass of about 0.9 Kg, pumps 10 litres / minute and as it is self contained, the patient can be mobile.

(<http://www.surgeons.org/asernip-s/net-s>).

Surgeons in the USA have implanted seven of these pumps. Only one patient was still alive after 12 months following implantation but later died due to strokes and other similar complications. After the pumps were removed, the atrial valve cage struts showed signs of thrombus and as result of these findings a change has been made to the design of the valve. This change is expected to improve the life expectancy of the patients.

Over this test, the average life span of 164 days was considered to be a good result considering that each patient was only expected to double his or her expected lifespan to around 60 days. (Abiomed, 2002 and Abio Cor lawsuit, 2002).

Table 2.8 is a summary of the clinical evaluation of the AbioCor pump up to the end of 2002.

Patient no.	Sex	Age	Where	Survival, days	Cause of death.
1	M	59	Jewish hospital	151	Strokes, bleeding.
2	M	69	Jewish hospital	391	Discharged, 27/3/02
3	M	-	Texas heart institute	144	Complications due to strokes.
4	M	74	University of California	56	Multi-organ failure
5	M	51	Hahnemann University	331	Discharged 22/1/02.
6	M	-	Texas Heart Institute	0	Bleeding
7	-	-	Jewish hospital	0	Unknown.

Takatani S., (2002)

Table 2.8. Summary of the clinical evaluation of the AbioCor totally implantable artificial heart. (TAH).

The exercise conducted with this TAH seems to be a very expensive way of finding that there are faults in the pump. From the figures presented, 5 out of 7 patients died from known causes and only 1 patient survived for about a year.

The failure of this exercise emphasises the danger of TAH application in that the patient is totally dependant on a man made device for his/her life. This is reinforced by Fox R. C., (1997), who indicated that the trials of the Jarvik-7 TAH were premature. If this pump type fails in any way, the life of the patient is at great risk as there no biological heart to fall back on.

2.2.11 Baylor pump.

This is a pulsating pump as a total heart replacement pump or total artificial heart, (TAH). The diameter and thickness of the pump are 97 and 82 mm respectively giving a volume of 510 ml. The pump has a mass of 0.62 kg and has two pumping chambers either side of an electromechanical actuator that consists of a DC electric motor and a roller screw mechanism. The rotary motion of the motor is converted into a linear motion by the roller screw, giving a pumping volume of 63 ml. During blood pumping, the roller screw pushes the pusher plate for blood ejection but it is not connected to the pusher plate during the blood filling phase. It is assumed the pump fills with blood passively. Sensors detect the position of the left pump to control the motor speed. Polyurethane or polyolefin rubber is used as the diaphragm material.

The TAH can give an output of 3-8 litres/min against an afterload of 100 mm Hg with a filling pressure of 1 – 10 mm Hg. (The term 'afterload' is the pressure that the TAH is pushing against when pumping blood out of the TAH). The power consumption ranged from 7 -12 W with an efficiency of 14% to 18%. (Takatani S., Shiono M., et al, 1992). The pump can be built as double assembly, as shown in Figure 2.11, as a TAH, or can be a single assembly as a Ventricle Assist Device, (VAD).

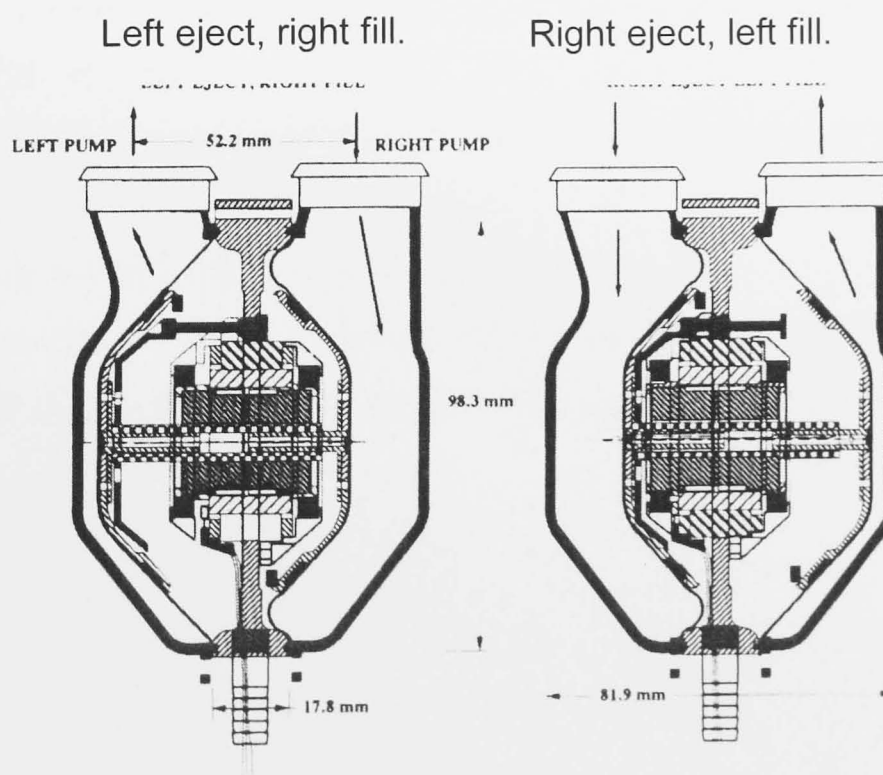


Figure 2.11. The Baylor total artificial heart. (TAH)

Hetzer, R., Hennig, E., Loebe, M. (editors), (1997)

2.2.12 The Novacor pump.

The Novacor pump is intended to be a fully implanted blood pump to assist patients with advanced heart failure. Work on the pump development started in the 1970's and progressed until in 1999 when it was claimed that 1000 patients had received this pump in one of the intended applications. It was also stated that one patient had been supported for 4 years on this pump.

The pump is a ventricle assist device, (VAD), and it is not intended to be a total artificial heart, (TAH). The pump is used in any one of three applications,

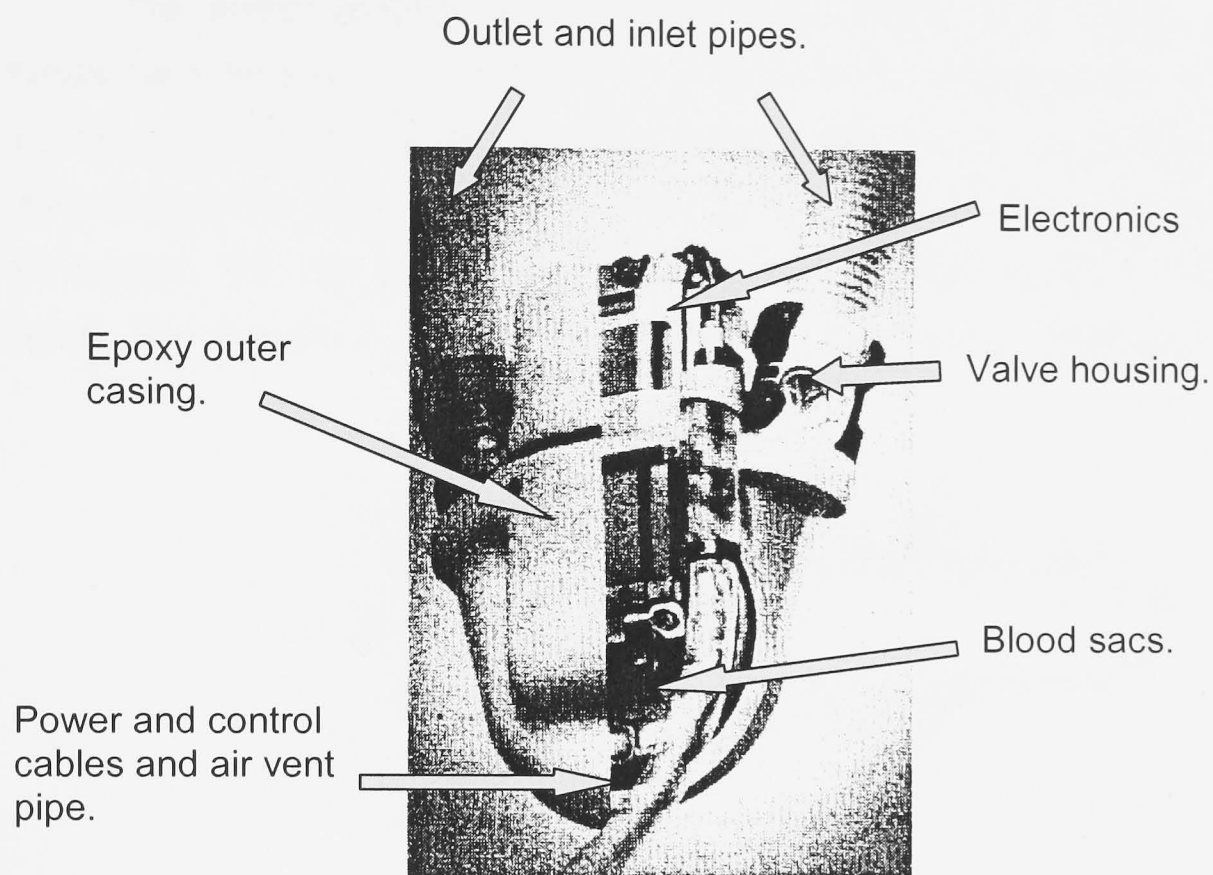
a) A bridge to transplantation. The pump is installed and is retained until a suitable donor heart is found for the patient. The pump is then replaced by the donor heart.

b) A bridge to recovery. The pump is implanted and assists the heart until the heart has recovered enough to be capable of operating unassisted. The pump is then removed.

c) An alternative to transplantation. The pump is implanted and left in position as a permanent fixture.

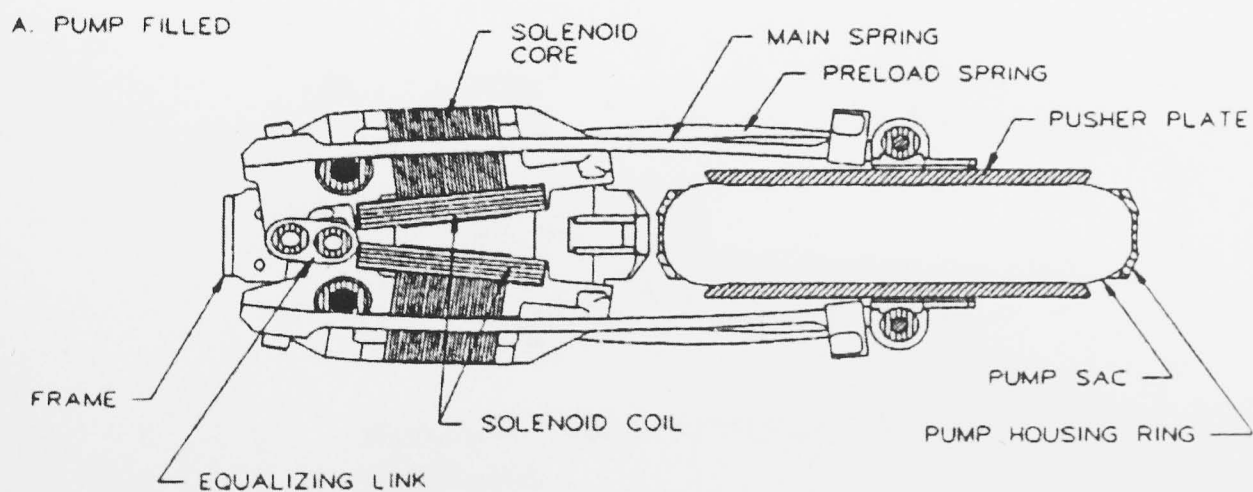
Figure 2.12 shows the general configuration and cross section of the pump which has a stroke volume of 70 ml and will pump 8.5 l/min at a preload pressure of 20 mm Hg.

The pump is implanted in a pocket made just below the ribs on the left side of the body. The pumped blood circuit is the normal arrangement for this application where the pump inlet is taken from the left ventricle and output is pumped back into the aorta. A full description is supplied by Wheeldon, D.R. et al., (2000). Figure 2.13 shows the assembly of the Novacor pump.



Wheeldon, (2000), Perfusion 2000; 15: 355-361

Figure 2.12 The Novacor pump, part section.



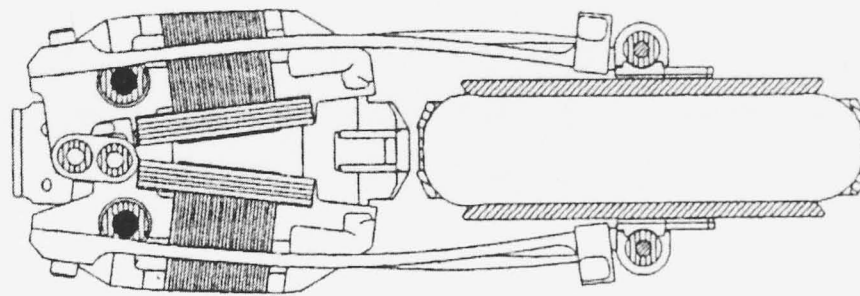
Pump Components

Wheeldon, (2000), Perfusion 2000; 15: 355-361

Figure 2.13. Assembly of Novacor pump.

2.2.13 Novacor pump operation.

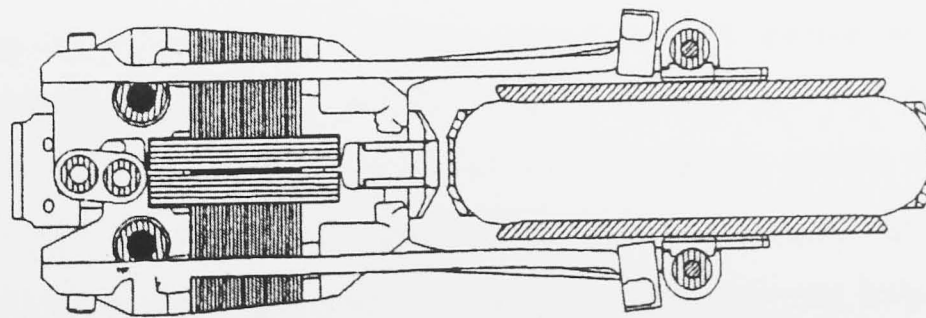
The pump consists of two solenoids operating in a scissor action. When the solenoids are energised, they compress the pumping sac, with the spring system producing a relatively constant load on the sac. At the end of the pumping, the solenoids de-energise and the blood sac fills passively from the patient. The sequence of operations is shown in the Figures 2.14a) to c) below. The direction of blood flow is controlled by valves built into the pump.



Pump full at end of filling phase

Wheeldon, (2000), *Perfusion* 2000; 15: 355-361

Figure 2.14a. The pump at the end of the filling stroke. The pump sac is full, the solenoids have not been activated and the springs are not flexed.

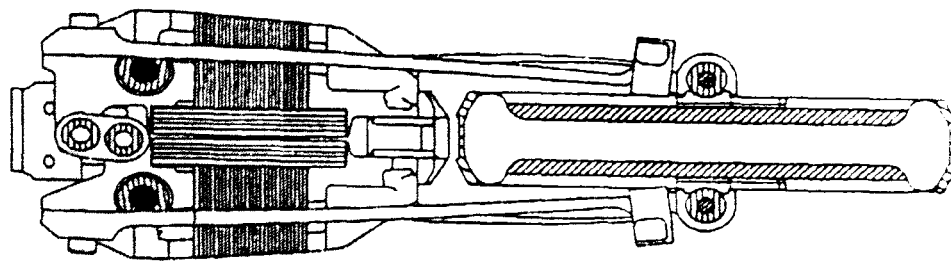


Solenoids closed, springs flexed, start of ejection.

Wheeldon, (2000), *Perfusion* 2000; 15: 355-361

Figure 2.14b. The pump at the start of the pumping stroke.

The solenoids are closed, the springs are flexed but the blood sac has not started to empty.



Pump effectively empty at the end of the eject phase.

Wheeldon, (2000), Perfusion 2000; 15: 355-361

Figure 2.14c The pump at the end of the pumping stroke.

The solenoids are closed and pressure from the springs has effectively emptied the blood pumping sac.

During pumping, the position of the moving parts is monitored by proximity sensors built into the pump.

The pump is electrically powered and an air vent pipe relieves the volume created by the pump operation with both cable and pipe passing through the patient abdomen. The pump can run in four modes, a) fixed rate, b) ECG triggered, c) fill rate triggered and d) fill-to-empty. All modes are automatically responsive to demand except for fixed rate.

There are no details of how the pumping sacs are connected to the valves. There could be problems if both sets of springs are not perfectly matched as there is likely to be axial movement of the sac as the stronger spring moves the sac towards the weaker spring. This may tear or rupture the flexible connections to the blood pipes. The blood sac may not fully empty under these conditions with all the associated problems with blood coagulation.

A failure point may be at the junction of the two solenoids. This is a mechanical junction and will be subject to loading every time the pump operates. Suitable bearing materials and design are necessary to ensure satisfactory operation for the time required for this application. It is assumed that operating parts of the pump are sealed from body fluids.

2.2.14 Spiral vortex pump.

The Spiral Vortex pump employs a 'wandering vortex' to give a wash out effect inside the pumping chamber. This pump developed by the Australian Artificial Heart Program is described by Umezu, M. (1994) as a pulsatile pump where blood enters tangentially and follows the downward motion of the diaphragm in a spiral motion. When the diaphragm rises, this spiral motion continues to circulate, moving upwards, blood washing the internal surfaces, until it is ejected from the pump.

This pump is claimed to have a lower index of haemolysis, at 32 – 35%, than a roller pump and a conventional pulsatile pump running under the same conditions. At an output of 3 l/min., the pump is not sufficient to fully support an adult but may be enough for a child. No details of applications were available. It was not stated what type of valves were fitted or how the pump was driven.

By design, pulsating pumps use valves to control the direction of the blood flow. Valves have been the cause of problems with the build up of calcified areas, pannus, clot formation and mechanical failure. Tilting disc, ball-in-cage types, animal tissue and artificial materials have been used, and are a topic of research by themselves.

The problems with valves will not be discussed here.

2.2.15 Other pulsatile pump types.

This section shows proposed mechanical circulatory support systems under investigation and demonstrates the variety of designs proposed. Following the time taken to develop previous pumps, if these pumps are successful, they are not expected to reach general use until well into the 21st century.

2.2.16 Radioisotope powered pump.

One Russian system proposed the use of plutonium 238 as fuel for a radioisotope powered, closed circuit, steam engine driving two ventricles. Steam at a temperature of 140° C and a pressure of approx. 3.5 kg/cm² is the working fluid for the steam engine. The use of tritium is proposed to eliminate

the hazard of radiation injury to the organs but produces other problems. There was no published information on how the pump output would be controlled. Figure 2.15 shows a general assembly of the pump. (Shumakov, Griaznov, 1983). It is likely that the use of implanted capsules containing plutonium 238 would not be accepted by the public.

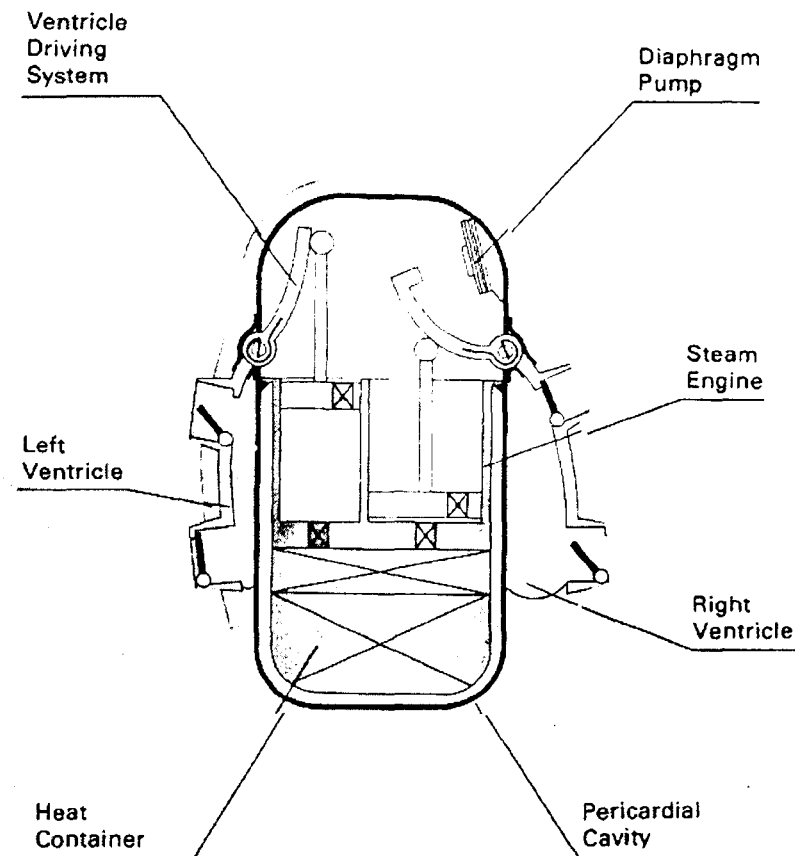


Figure 2.15. Radioisotope powered heart pump

Shumakov, Griaznov et al., *Artificial Organs*, 7(1): pp101-106, (1983)

2.2.17 The pendulum pump.

In this Korean design, Figure 2.16, the motor moves back and forth, similar to a pendulum, using an epicyclic gear mechanism. This motion alternately compresses two bladders, one for each ventricle, with valves controlling the direction of blood flow. It is not necessary to use a compliance chamber as moving the motor instead of pusher plates saves the dead space taken up by the motor.

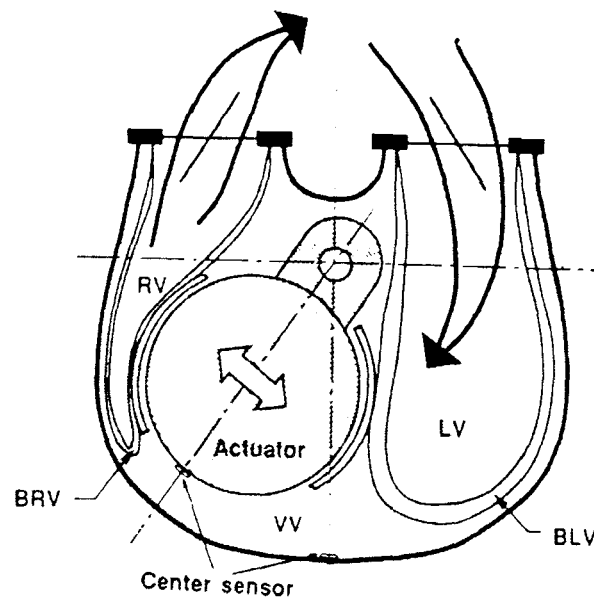


Figure 2.16. Korean pendulum pump.

Hetzer, R., Hennig, E., Loebe, M. (editors), (1997)

There is no information on the performance of this pump in terms of flows and pressures generated.

It is a major problem to ensure that there is a suitable 'washout' of the bladder after each stroke as any blood left may react with the bladder material and cause blood clots. If these clots become detached they could pass to the patient. Another problem is movement of the pendulum against the bladder may cause wear / puncture of the bladder material.

2.2.18 Aachen total artificial heart.

A pump by the Helmholtz Institute at Aachen, used a similar action to the Korean pendulum pump to drive two pusher plates, compressing and releasing two diaphragms with leaflet valves controlling blood flow. The pump, Figure 2.17, shows the pusher plates, 4, pivoted at X, driven by the crank mechanism, 10, to alternately compress and release the two bladders or diaphragms, 6, that pump the blood through the valves, 1. The crank is driven by an electric motor, 8, through reduction gears, 9. The vent, 11, allows the power and control cables to enter the pump.

This pump does not operate the same as the human heart as each ventricle will pump alternately. This effect on the body is unknown at present.

Both the pumping chambers must not have regions where blood can pool or stagnate leading to thrombus problems.

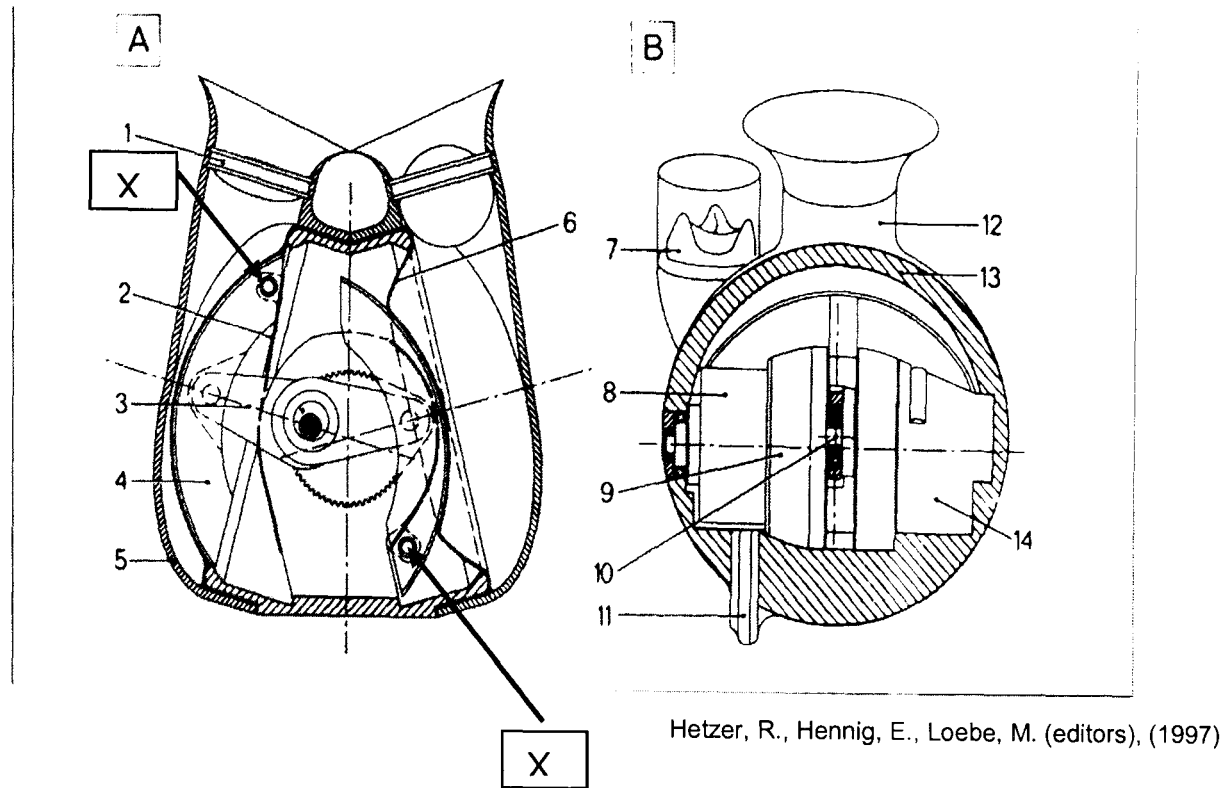


Figure 2.17, General arrangement of the Aachen total artificial heart.

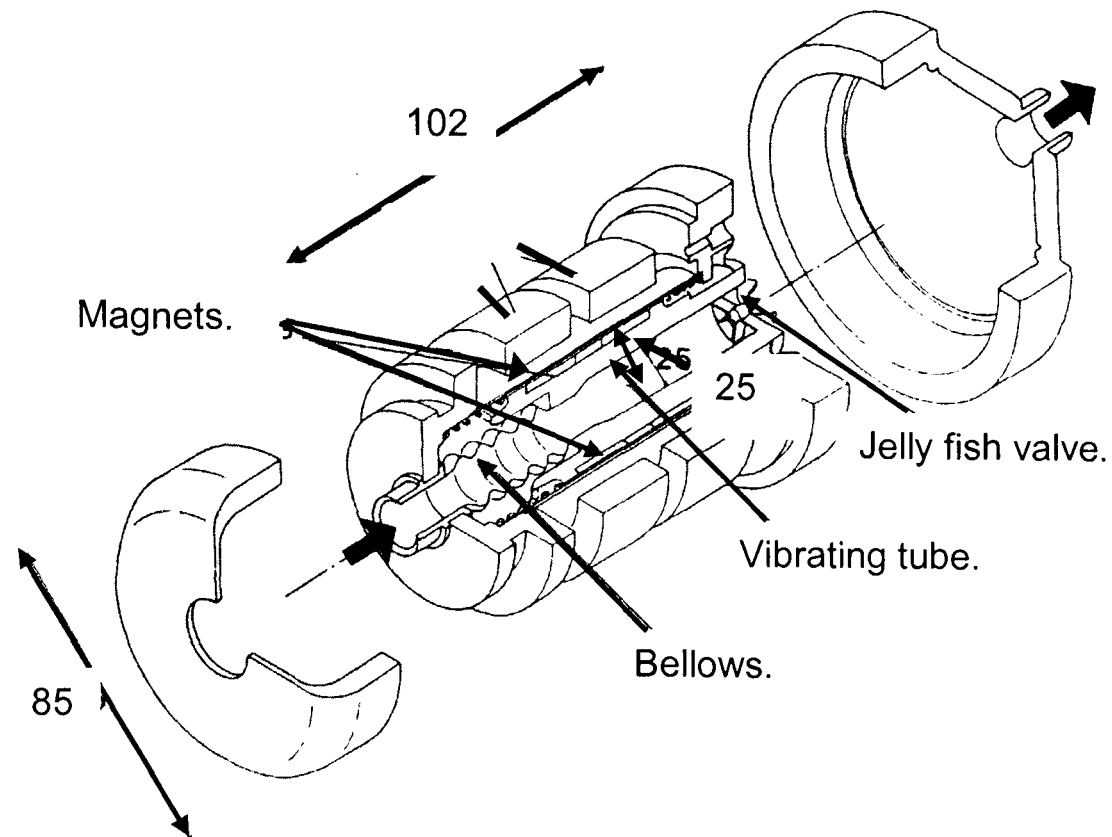
The pusher plates are pivoted at point X. As the plates move, there may be relative movement between the plate and the non blood surface of the blood sac. This may cause mechanical wear which could lead to premature failure of the sac and is similar to the Korean pendulum pump previously described.

2.2.19 Vibrating flow pump (1).

This pump, designed at Tohoku University, Japan, is a left ventricle assist system. It is described by Kobayashi S., Nitta S., et al. (1998), as an elastic tube, with magnets attached to the outside of the tube. Figure 2.18 details the assembly. The tube vibrates in an electromagnetic field at 10 – 50 Hz and blood enters at one end of the pump and is ejected at the other, the blood direction being controlled by a jelly fish valve at the outlet end.

At 25 Hz the pump delivery is 2 l/min. This is not enough to maintain life as a heart replacement pump but may be satisfactory as a child pump or adult assisting pump. It was stated that *'there was no fatal arrhythmia observed during this study'*. Comments by Kawano S., Isoyama T., et al,

(2003), indicate that *'the magnets made the pump heavy and the vibration system generated an amount of heat'*.



Hetzer, R., Hennig, E., Loebe, M. (editors), (1997)

Figure 2.18. Exploded assembly of the vibrating flow pump.

Sizes are in millimetres.

2.2.20 Vibrating flow pump. (2)

Similar to previous work, another vibrating flow pump has been developed at Tohoku University, Japan, by Kawano S., Isoyama T., et al, (2003). This pump uses a cross slider mechanism, (a Scotch Yoke), which converts rotary motion into reciprocating, simple harmonic, linear motion. The smoothness and life of this mechanism depends on material quality and pump speed. Higher speeds will give greater wear and consequent deterioration in the quality of the movement.

The vibrating tube has an inner diameter 4 mm and a vibration amplitude of 1.5 mm. The total weight of motor and pump is 50 grams. Bellows allow the vibrating part to move and to connect to the rest of the pump with a jellyfish valve controlling the direction of blood flow. The motor is some distance from the vibrating tube and there is no heat transfer to the blood.

With a tube frequency of 50 Hz, the maximum pressure achieved is about 60 mm Hg with a flow rate of 0.06 l/min. Figure 2.19 shows the general assembly of the pump.

The output of this pump is not enough as a left ventricle assist device but could be suitable as an external shunt pump to provide additional blood to coronary arteries.

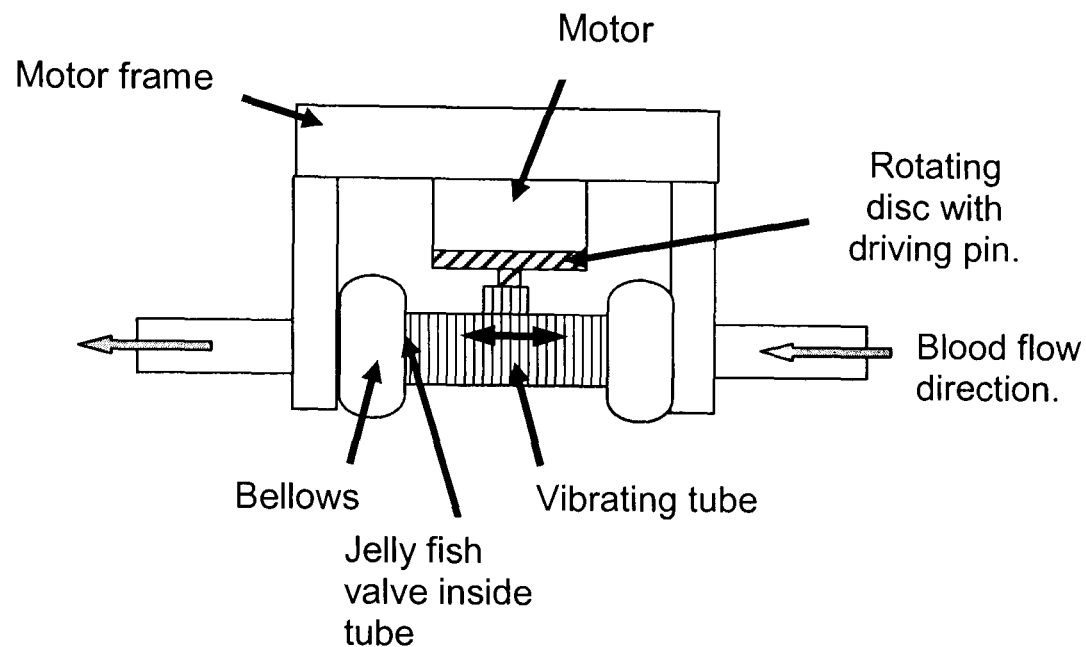


Figure 2.19 General assembly of the vibrating flow blood pump.

2.2.21 Muscle activated pumps.

Attempts have been made to develop blood pumps powered by skeletal muscle. This is different from heart muscle, it fatigues quickly and is not suited to long term operation as required for a blood pump.

Salmons S., Vrbová G, et al., (1969) and Salmons S., Sréter F., et al., (1976), found that a conditioning process can change skeletal muscle to muscle more suited to long term function. Acker M., Hammond R., et al., (1987), indicate that it can be made to be so fatigue resistant that it can be used for cardiac assistance.

A schematic layout of a muscle powered blood pump has been shown previously, Figure 2.1. This shows a stimulator causes muscle to contract. The muscle contraction moves a plunger which pumps fluid into a hydraulic actuator. The actuator moves a diaphragm, which pumps the blood.

For an enlarged heart condition, (dilated cardiomyopathy), skeletal muscle is wrapped around the heart and, using a pacemaker, squeezes the enlarged heart to enable improved pumping. (Pierce W., 2000). Another proposal wraps the muscle around the aorta, after the coronary arteries, and with timed muscle contractions, improves the blood flow to the coronary arteries. Tests on dogs by Thomas and Isoda, et al, (1996), and Thomas and Hammond, et al, (1999), have shown this procedure to be possible for up to 4 years. This assistance appears to be similar to the intra aortic balloon pump method, where extra blood is forced into the coronary arteries by blocking the aorta at correctly timed intervals.

Other work with this type of power source are given by Farrar, Reichenbach et al, (1995), Takatani, Takami et al, (1995) and Mizuhara, Koshiji et al., (1995).

These systems are not as advanced and have not had the success as the alternative pumping systems. Work is continuing to evaluate alternative power sources. These will not be discussed here.

2.2.22 Summary of pulsating pumps.

Pulsating pumps are the first generation of blood pumps. Some have reached the production stage; some are still in development while others are unlikely to proceed any further.

With pulsating pumps, the length of time the blood is inside the pump is important. The greater the surface contact time, the greater the chance of interactions between the blood and the materials. Reul H. and Akdis M., (2000), indicate that the blood 'dwell' time for a pulsating pump can be greater than 1 second.

The pumping action of a pulsating pump has to be carefully managed as it is possible to create 'dead' areas inside the pumping chamber where the blood is stationary or in a stationary vortex. This may lead to blood clots.

Unsatisfactory pumping actions can create shear stresses causing adverse reactions in the blood. Tissue growth into valves or pump outlets can cause problems in terms of reduced flow or regurgitation.

A compliance chamber or other means of venting the volume created and destroyed under the pumping diaphragm must be made. The pneumatic HeartMate pump uses this volume to operate the pump but both the electric and pneumatic versions use a pipe, or cable, passing through the skin to operate or ventilate the pump. Comments regarding the penetration of the skin are made later in this summary. Other systems use a small bladder installed inside the patient. The bladder connects to the volume on the non blood side of the diaphragm and acts as a compliance chamber. If the bladder is faulty, a surgical operation may be necessary with the associated discomfort to the patient.

The diaphragm of a pulsating pump is the major component. Extreme care is necessary to ensure that it has a satisfactory fatigue life and that installation does not cause folding or crumpling that could lead to diaphragm failure or thrombus creation.

With an internally fitted pulsating pump, the patient may experience vibration. The vibration, or noise, may not be acceptable to the patient.

The pump must be attached to a 'structural' part of the patient as leaving the pump hanging from the pipes or cables, arteries or veins must be very risky. The pump must be securely 'installed' to the patient, otherwise it may move as the patient moves. Installing the pump in a 'pocket' constructed by the surgeon appears to be the present method but leaves the pump liable to start tissue reactions.

The power supply to the pump has received some attention. Pipes and cables passing through the skin are not satisfactory as it leaves the patient open to infections. A system to transfer power through the skin is the best that can be achieved at present and is described later.

The problem of battery life needs to be addressed. A life of 8 hours is just about long enough for overnight but must restrict the life of the patient in terms of daily living, work and social activities. A battery with a greater power density would be of great benefit enabling the patient to live a near normal life.

The problem of producing a completely satisfactory artificial heart has not been solved. Even with some of the latest designs, patients are still dying prematurely.

The next part of this chapter covers the pumps that are non pulsating types. These are rotary pumps that do not use valves and provide a constant, non-pulsating, blood flow.

2.3 Continuous flow pumps.

2.3.1 Introduction

A continuous flow pump can run at a fixed or variable speed and has the advantages over a pulsating type pump of full implantability, a significant reduction in physical size, no valves, no diaphragms and the potential for higher reliability and longer durability. A continuous flow pump will not have a limited fill capacity or require a compliance chamber.

There is no long term experience of the effect of non pulsatile blood supply to the organs of the body although, the pulse effect is less further away from the heart, due to the combination of the elasticity and reduction in size of the arteries and the viscosity of the blood.

Reul, H and Akdis, M, (2000) indicate that blood can stay inside a pulsating pump for greater than 1s but with a constant flow pump it can be less, (1s – 1 ms). Therefore the chance of creating blood clots due to surface interactions with a constant flow pump is reduced compared to a pulsating pump.

A disadvantage of this type of pump is with bearings and seals, if these are blood immersed, they have the potential to cause problems. Possible solutions will be discussed later in this chapter and will include mechanical type bearings and magnetic suspension systems.

2.3.2 Radial flow pumps.

With a radial flow pump, the fluid enters in line with the rotating drive shaft of the pump. The fluid is collected by a rotating disc(s) and leaves the pump tangentially through a delivery outlet. A magnetic drive or electric motor drives the pump.

Many continuous flow blood pumps imitated existing hydraulic pumps with bladed discs but were found to cause problems with blood clots, cavitation and resulted in patient problems. Sealing the driving shaft against blood leakage without blood damage has been a problem. Rotary shaft seals are suitable in the short term, (hours), but are prone to leak or fail with blood leakage. Industrial sealing methods are not suitable for blood pumps. One option appears to be blood lubricated bearings. These have problems which will be described later.

Radial flow pumps can run with a greater clearance between the blade, or rotor tip, and the pump casing than axial flow pumps without serious effects on pump efficiency. Jarvik R., (1995), indicated that a radial clearance of 0.102 mm was practical for an axial flow pump but did not offer any figures for a radial flow pump.

The possibility of increased rotor clearance makes magnetic suspension bearing systems more suitable for a blood pump as there is less danger of the rotor colliding with the pump casing and these would seem to be the ideal at present.

The clearance of blade tips relative to the inside of the pump casing is a concern in terms of the shear stress to the blood. This will probably be the governing consideration as it may be better to sacrifice an amount of efficiency but have a 'safer' pump in terms of haemolysis.

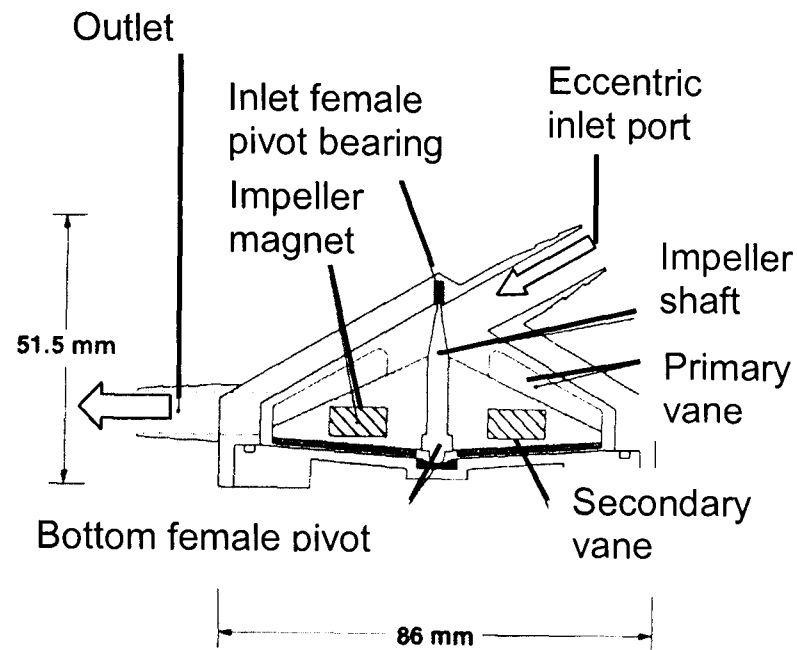
Examples of radial flow pumps follow.

2.3.3 Baylor rotary pump.

Figure 2.20 outlines the general arrangement of the Baylor rotary radial flow pump. The part shown is the blood pumping part and does not show the pump driving arrangements.

This pump is intended to be suitable for extracorporeal and long term intracorporeal ventricular assistance.

A brief description of this pump is taken from Ohara, Makinouchi, et al., (1994) and is summarised below.



Hetzer, R., Hennig, E., Loebe, M. (editors), (1997) and
Ohara Y, Makinouchi K., et al. (1994)

Figure 2.20, The Baylor rotary pump.

Previous pumps using a drive shaft and seal arrangement suffered from blood leakage and the pump needed to be changed every 24 hrs for trouble free running. The pump was unsuitable for implantation and a different method of mounting the rotor was required.

With a later design the blood enters through an eccentric inlet port and is 'collected' by vanes on the rotor. The blood leaves the pump through the tangential outlet port. It is assumed that the small secondary vanes on the back of the rotor give a 'wash out' effect as this is a possible location for blood clots. As the inlet port is eccentric, the top bearing of the rotor is out of the blood flow as experience had shown that bearing support bar(s) across the pump entry may cause thrombus formation.

The rotor is mounted on two small pivot bearings. Careful design of the bearings is necessary in order to minimise blood damage and this design indicates that the bearings are similar to a mechanical clock where a small tapered shaft fits into a cup. This arrangement gives both axial and radial

location with a small contact area. The bearings used in this pump were made from carbon steel and brass which are not good choices for blood contacting materials. After 120 hours of *in-vitro* tests, wear of '*less than 0.2 mm*' was reported. This is high for an implanted pump and more work needs to be done to solve this problem for a target life of 3 months.

Yamane T., et al., (2004), estimated that the minimum shear rate around the bottom pivot was 650/s '*which is higher than the threshold for thrombus formation*'.

The rotor is driven by internal magnets that follow the rotating magnetic field provided by electronics external to the pump. (The electronics are not shown).

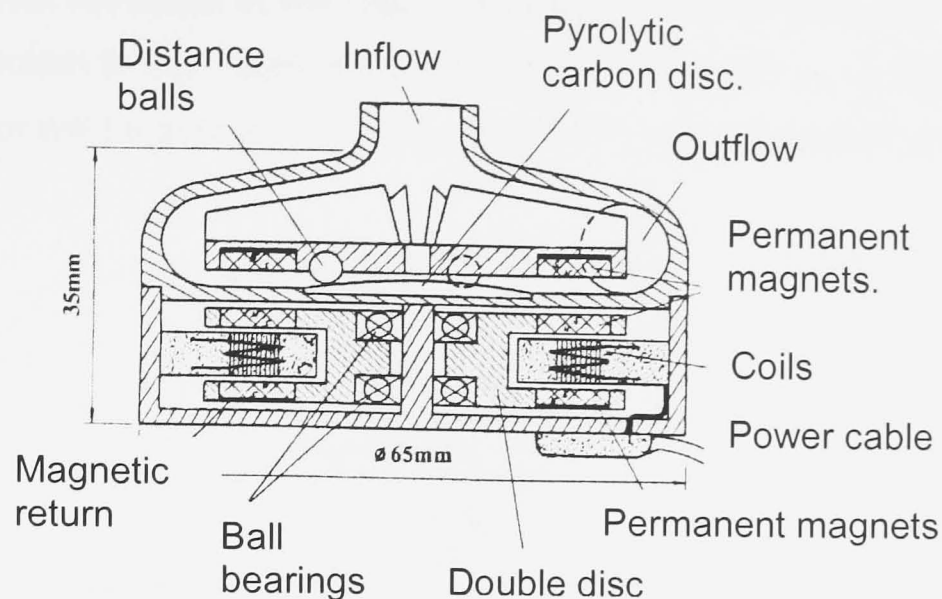
The pump can deliver 8 l/min at a pressure of 250 mm Hg at a speed of 2400 rev/min. At the same speed, and a delivery of 1 l/min, the pressure rise is greater than 300 mm Hg.

2.3.4 Vienna pump.

The description of this pump is taken from Hetzer, R. et al., (editors), (1997). This is a rotary pump with inbuilt electronics producing a flow of 10 l/min at a pressure head of 100 mm Hg. with an overall efficiency of 11% at 2,500 rev/min.

Figure 2.21 shows the general layout. The pump has a diameter of 0.065 m, a volume of 101 ml and a mass of 0.240 kg. and is built with a blood pumping section and a section containing the electronics and magnets that drive the pump.

The pumping rotor appears to be supported on three ball bearings, running on a convex shaped carbon disc, that are immersed in blood. This arrangement is likely to cause damage to the blood cells. Although blood contact with the carbon disc itself is unlikely to cause haemolysis, it is expected that the action of the balls running over the surface of the disc will mechanically damage blood cells. During evaluation tests it was noted '*some blood trauma was caused*' with a '*haemolysis index as low as 0.0046 g /100 L*'.



Hetzer, R., Hennig, E., Loebe, M. (editors), (1997)

Figure 2.21 The Vienna centrifugal pump.

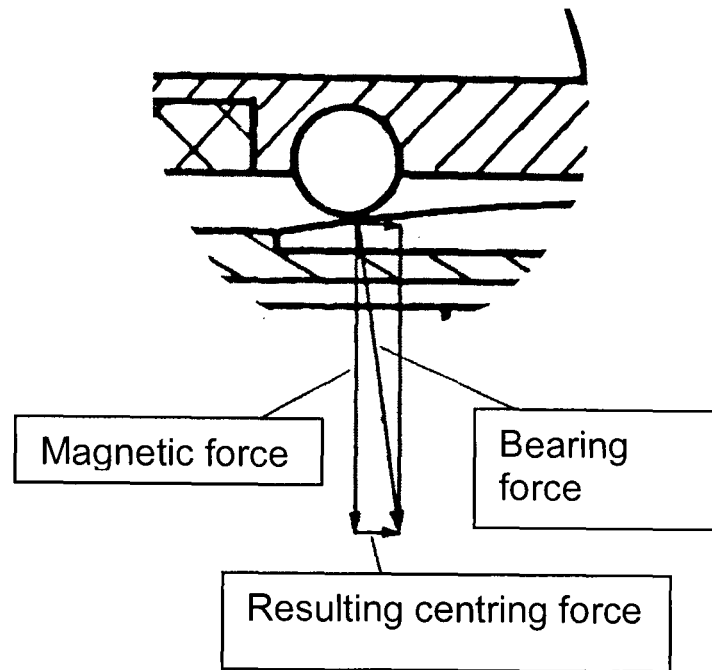
During testing, depending on the strength of the magnets used, the rotation of the rotor on the three ball bearings was stable up to a delivery of 10 l/min at a pressure of 200 mmHg. Beyond this point the rotor tended to lift from the carbon disc and run 'off centre' touching the inside of the housing. Stronger magnets could be used but would mean greater contact forces at lower speeds.

It is advantageous to run the pump at a speed and load so that the rotor is just on the point of lifting off the surface of the carbon disc in order to minimise haemolysis and prolong the pump life. There is nothing to show if this was attempted. Schima H., et al., (1995) reported small amounts of thrombi were found around the area where the distance balls were located but this was considered to be a manufacturing fault.

Figure 2.22 indicates that if the rotor runs 'off centre', the resultant forces will pull the rotor back to its correct position.

The vector shown as 'magnetic force' is understood in terms of direction and magnitude. The direction of the vectors shown as 'bearing force' and 'centering force' are not understood. As shown the ball has two downward forces with no equal and opposite reaction forces. Any reaction force will be 'normal' to the disc surface and will resolve into a vertical force opposing the magnetic attraction, and a horizontal force tending to make the

ball move down the slope of the disc. The horizontal force vector will be in the opposite direction to the 'resulting centring force' as shown as on Figure 2.22, i.e., the vector will be pulling outwards and will not be a centralising force.



Schima H., Schmallegger H. et al., (1995)

Figure 2.22 Diagram showing the published forces.

With the three balls running on the carbon disc, equally spaced around the circumference and assuming all are experiencing identical forces, the rotor will be in unstable equilibrium. If the rotor moves 'off centre' by a small amount, one horizontal force will increase while the others will decrease as the 'off centre' bearing ball moves down the slope of the carbon disc. As the rotor moves 'off centre', the magnetic attraction between the magnets in the rotor and one, or more, of the permanent magnets in the base of the pump will become greater as one the rotor magnets becomes closer to the base magnets. This will result in the so called 'centralising' force on the 'off centre' side of the rotor becoming larger while the other will become smaller. The rotor will then run eccentrically. This condition is not self correcting and the rotor will possibly collide with the pump casing.

The pumping rotor appears to have a hole in the centre. This may be an attempt to prevent thrombus formation under the disc as there will be a 'wash out' effect under the disc as blood passing through the hole to the underside of the disc will be drawn around with it and move outwards with the centrifugal forces created by disc rotation.

2.3.5 Blood damage with rolling bearings.

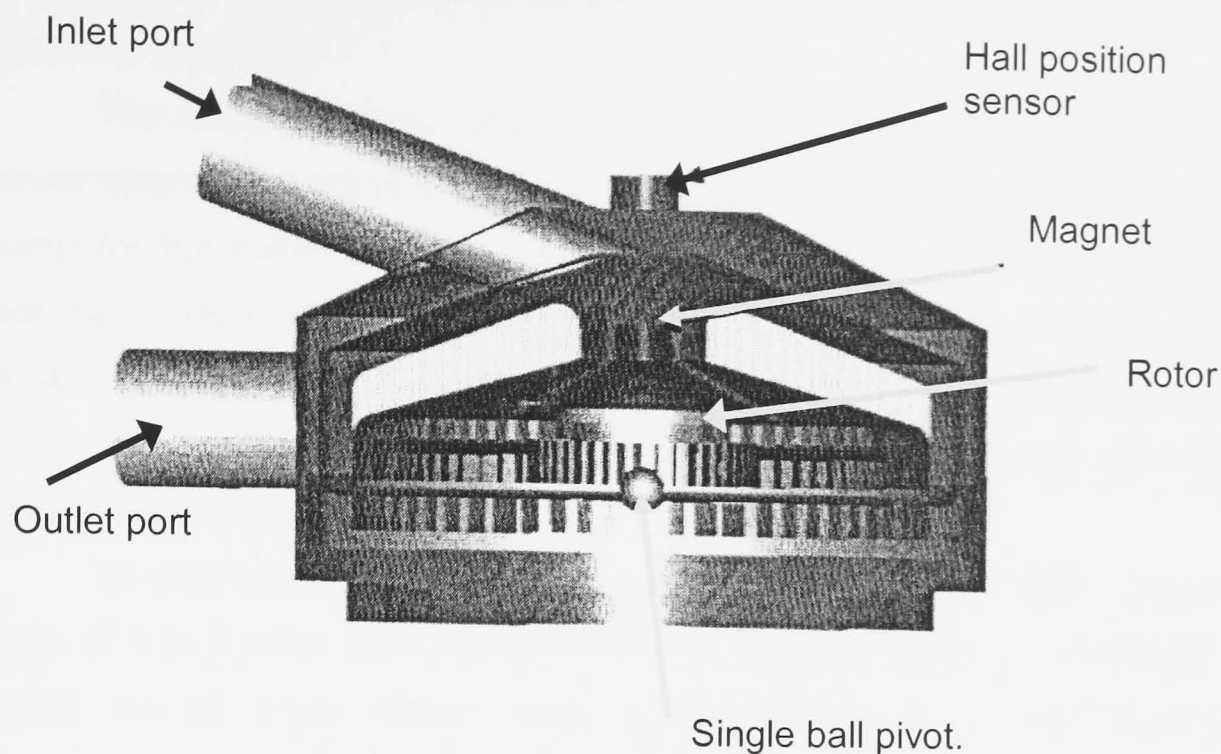
Following the description of the Vienna pump, a literature search was made concerning blood damage with rolling bearings.

Fok and Schubotho, (1960), used a flask, containing quartz beads and blood, rotating on a horizontal axis. The results from this work indicated haemolysis was present and was related to the length of time of rotation and the number of beads in the flask. The paper suggested that haemolysis might be produced by high fluid stresses damaging the blood cells as the beads moved through the blood and not by direct physical crushing of the cells between the quartz bead and the wall of the flask.

Freitas R., (2003), indicates 'impact' haemolysis is not unknown in marathon runners and other physical sports where *'although non mechanical factors may contribute, it is generally believed that most of the damage is caused by mechanical tearing'*. An example is given of where *'a human runner, of weight $\sim 10^3$ N whose footfall force is spread over an area of $\sim 100\text{cm}^2$, can exert a momentary tissue pressure of $\sim 10^5$ N/m² which is well in excess of the red cell fragmentation stress limit of 150-250 N/m²'*.

2.3.6 Development of the Vienna pump.

Yoshino and Uemura et al., (2001) proposed a development of the Vienna pump. It had the same dimensions as before, outer diameter 60 mm, height 50 mm, mass 760 g, but the three ball bearing support system was replaced by a single ceramic ball mounted under the impeller. A polyethylene female pivot was built into the pump casing. No other means of supporting the mass of the rotor was used. A magnet at the top of the rotor was used to stabilise, or centralise, the rotor. Figure 2.23 shows the general arrangement.



Yoshino M. and Uemura M. et al. (2001.)

Figure 2.23. Vienna single pivot centrifugal blood pump.

At 1,200 rev/min the rotor tended to wobble but at 1,400 rev/min, the rotor was stable. At 1,800 rev/min the rotor started to lift from the bearing. Output was 5 l/min at 2,200 rev/min against back pressure of 100 mm Hg. Reducing back pressure to 52 mm Hg. gave 9 l/min. There was no information to indicate what would happen if the rotor lifted enough to lose contact with the ball pivot at the base of the rotor.

2.3.7 AB 180 pump.

Westaby S., (2000), describes the AB180 as a radial flow centrifugal pump constructed in two parts. The upper housing contains a 6 bladed 25mm diameter impeller, running at 2500 – 4500 rev/min., attached to a stainless steel rotor inside a lower housing which contains a stator system. The base of the rotor sits within a magnetic field. A three phase, 12V system uses two phases for power, the third phase is used for speed sensing.

Sterile water pumped into the lower housing provides a fluid bearing to the rotating surfaces and a heparin delivery system is used for blood anticoagulation.

A balloon mechanism is fitted to prevent any blood backflow in case the pump fails.

The microelectronic system contains a transformer, a battery to supply power when transporting the patient, water and heparin infusion pumps, an air pump for the balloon system, alarms and associated electronics. The only external control is a speed control knob. This implies a manual control system. The pump inflow is inserted into the left atrium with the pump outflow connected to the ascending aorta. This implies that the left ventricle is 'unloaded' and is rested.

'In vivo' tests using sheep were done by Clark. et al, (1998). Blood flow rates of 1 to 5 l/min were achieved over a cumulative time of 106 days. The pump ran at 4,162 rev/min and maintained 73.5% of the blood flow requirements of the animal(s) with normal indications for blood parameters. The size and weight of the implanted pump reduced the chances of infection.

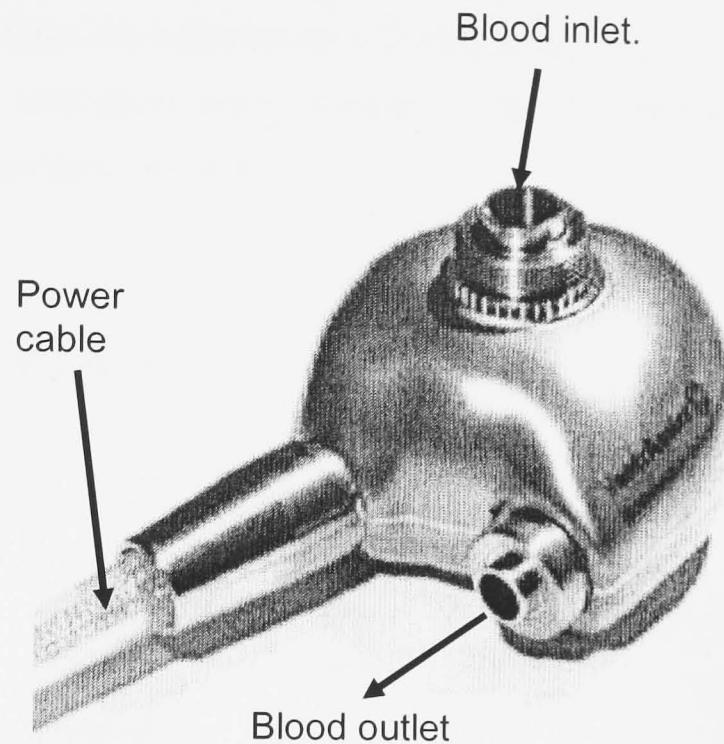
'In vivo' tests by Tevæearai et al., (2001), using calves of a weight similar to a human, (71 kg), indicated a delivery of 5.6 l/min at 4500 rev/min with acceptable blood pressures.

2.3.8 VentrAssist radial flow pump.

Peterson and Woods, (July 2004), report that this pump, made by the Australian company Ventracor, is designed as a left ventricular assist device. It only has one moving part, the impeller, that is suspended on the blood as it rotates. Coils in the pump casing generate the magnetic fields that are picked up by the magnets in the impeller. Suitable electronics enable the impeller to rotate without any other bearing support. The life of this pump is unknown as there are no contacting moving parts to wear.

The clearance between the rotor and casing and the 'stiffness' of the bearing system will be critical. Unless the pump is securely anchored inside the patient, are there gyroscopic effects likely to affect the patient?

The fully implantable titanium pump, shown in Figure 2.24, is 60 mm in diameter, weighs 0.298 kg with a volume of 122 ml. It is designed to be suitable for both children and adults with a maximum flow rate of 10 l/min.



Peterson I., Woods D., (July 2004).

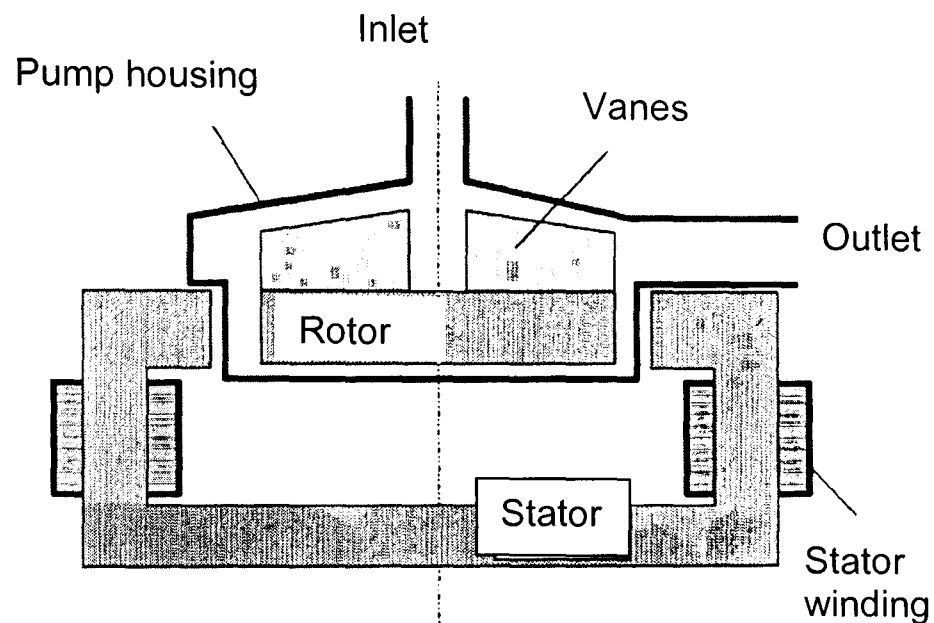
Fig 2.24, VentrAssist blood pump.

2.3.9 HeartMate III pump.

Schöb R., Loree H.M., (Date accessed 2005), describe this is a radial flow pump with a full magnetic suspension and drive system. Where the original diaphragm type pumps had an anticipated life of 2 – 3 years this system is intended to make progress towards an LVAD with a life of 10 – 15 years. The pump is in two parts, a blood pumping chamber, and a suspension and driving section. This enables the blood pumping section to be separate from the electronics and to be fully disposable. The blood pumping chamber is made from titanium and the internal surface is textured to promote the growth of the pseudoneointimal layer, similar to the HeartMate I pump.

The rotor weighs 0.349 kg and the magnetic suspension system can handle forces of 31.5 times this radially and 23.1 times this force axially. This gives a very robust suspension system.

In vitro tests gave a flow rate of 7 l/min at a pressure of 135 mmHg at a speed of 4,800 rev/min. *In vivo* tests over 27 to 61 days have given satisfactory results at flow rates of 3 to 11 l/min. with no pump failures and no blood problems. An examination of one pump, after 59 days, showed a blood compatible biological lining similar to the HeartMate I pump. Figure 2.25 shows the general assembly.



Schöb R., Loree H. (Date accessed 2005).

Figure 2.25, The general arrangement of the HeartMate III pump.

The magnetic suspension will give an infinite life assuming no collisions with the walls of the pumping chamber. Care will have to be taken with the control system as it will be more complicated than other systems. Two sub-systems run together so that in case of a fault, one is capable of running the pump without interruption. Suitable reporting systems have been built in to signify any faults.

The magnetic levitation system appears to be an answer to the problem of running a blood pumping device. One problem has not been answered however. Blood contains haemoglobin which contains iron. The effects of strong magnetic fields, (up to 2 Tesla), on haematology have been studied and no consistent results have been found. It is assumed that the

amount of magnetism involved with a magnetically driven and controlled blood pump will be small and the effects will be negligible, (Moulder J., 2001, and Moulder J., 2004).

2.3.10 Description of axial flow pumps.

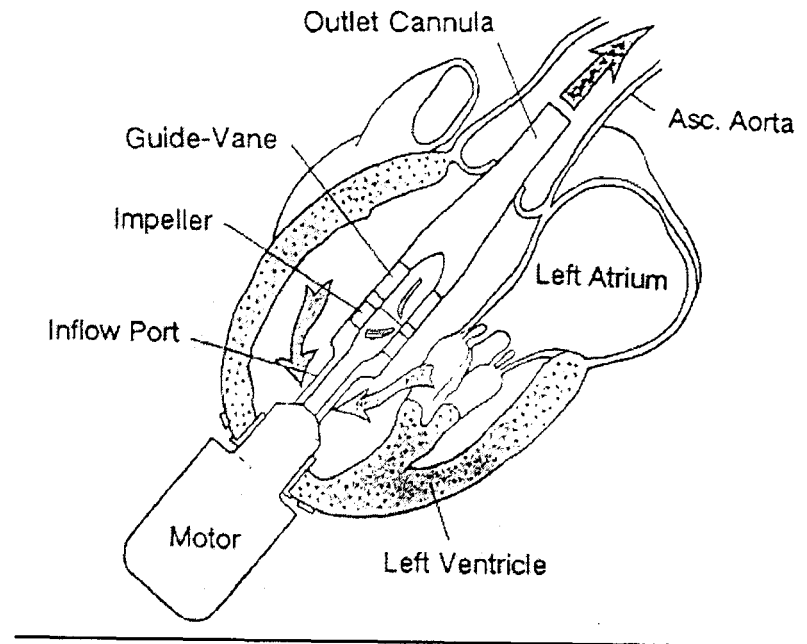
In axial flow pump, fluid is drawn in at one end and exits at the other end along the line of the central rotating shaft and consists of a circular section barrel with a screw or propeller device mounted on bearings inside the barrel. The pumping action is created when the screw rotates and fluid moves axially through the barrel.

Reul and Adkis, (2000), show that the rotation speeds can range from 6,000 to 160,000 rev/min. Bearing types include blood lubricated ceramic bearings, diamond coatings, ball bearings and magnetically levitated bearings.

2.3.11 Intra ventricular axial flow pump.

This pump type described by Yamane, Ikeda, et al., (1996), is an axial flow pump designed to draw blood from the inside the ventricle and pump it directly into the ascending aorta. Power is supplied to the motor attached to and external to the ventricle. Figure 2.26 shows an installation of a pump of this type where it is sewn to the outside of the left ventricle. The pump is powered using an electrical cable passing through the skin.

Care must be taken to ensure that no part of the ventricle is drawn into the pump inlet. Apart from damage to the heart, the blood flow will be interrupted and may stop even though the motor is still running.



Yamane, T. Ikeda, T et al, (1996).

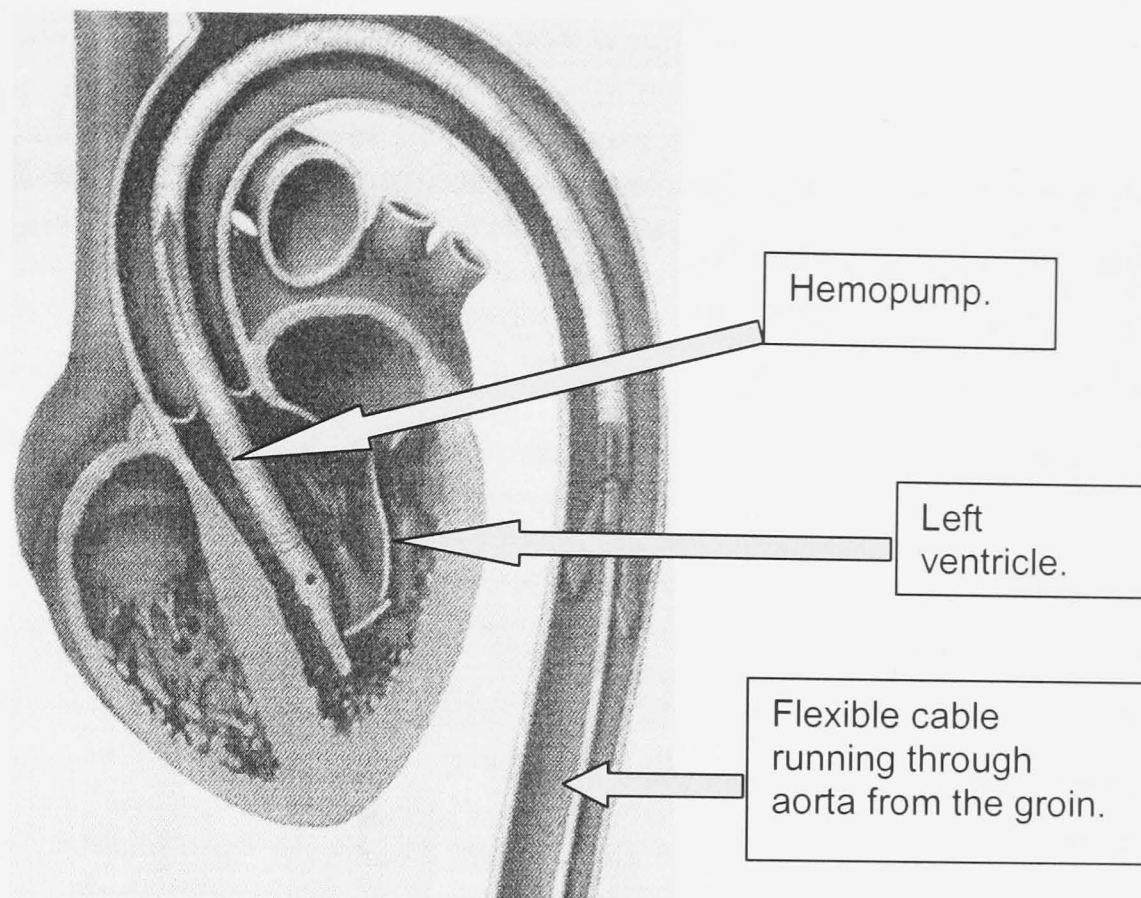
Figure 2.26. Installation of intraventricular axial flow pump.

2.3.12 Hemopump.

Lönn, U., (1997), shows that the Hemopump is inserted into the aorta in the groin and passed up to the heart. The pump is positioned in the left ventricle and pumps blood from the interior of the left ventricle into the aorta. Figure 2.27 shows this arrangement. It is noted that the pump drive passes through the aortic valve and that the valve is held open while the pump is in position. There is no mention of damage to the valve.

The best methods of driving these pumps so far include rotating magnetic field arrangements.

Axial flow pumps have the advantages of being fully implantable, they are small and quiet, there are no valves required, and by varying the speed they can be pulsatile or non – pulsatile and are less complex than pulsatile pumps. They do not need a compliance chamber or atmospheric vent system. This reduces infection due to pipes passing through the skin. (Frazier, 2003). With non pulsatile support, the blood ejection pulse by the patient's heart on the arterial pressure trace can be seen. This can be used as an indicator of the performance of the patient's heart. This trace is difficult to see with pulsatile support. (Dabritz and Messmer, 1997)



Lönn, U. (1997).

Figure 2.27. The installation of the Hemopump.

A disadvantage with any pump is if the internal 'hydraulic' conditions are not suitable for pumping blood safely and cavitation, high instantaneous pressures and high shear forces are present, blood damage could occur causing problems to the patient.

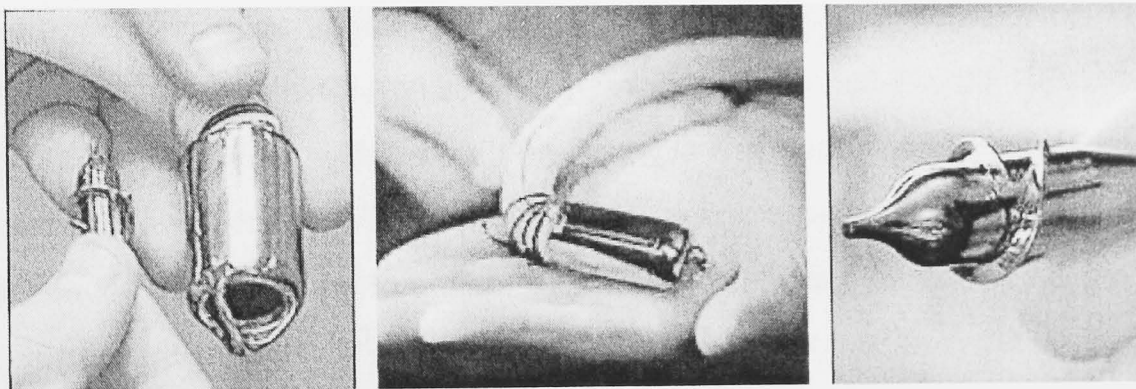
The exact safe levels of blood shear are difficult to determine. Some investigators report platelet activation and adherence to blood vessel walls can increase with high shear rates, (Markou, Hanson, et al., 1993). Others suggest that the blood vessel walls respond to the changes in wall shear rate rather than the level of shear, (Frangos, Eskin, 1985). Some evidence exists that atherosclerotic thickening and intimal hyperplasia occur at regions where the average wall shear stress is low and is oscillatory, (He and Ku, 1996 and Ku D. 1997 and Wotton D.W. and Ku D.N., 1999). This is an advantage for a non-pulsatile pump. The arteries appear to adapt to a 'normal' wall shear stress average of 1.5 to 1.0 N/m², where a consistently high shear stress causes dilation, (enlargement), and a consistently low shear stress causes them to reduce in size. (Glagov and Zarins, et al. 1988).

2.3.13 Jarvik 2000

One of the later, more successful artificial blood pumps, the Jarvik 2000, shown in Figure 2.28, is a non-pulsating, continuous flow, axial pump. The fully implanted pump measures 0.025 m in diameter, has a mass of about 90 grams and has a displacement volume of 25 ml. Animal experiments indicated that the pump was free from thrombus problems. (Fuchs, Netz, 2002).

The pump has a mean power consumption of 6 – 7 watts and the 12V batteries last about eight hours. The pump speed, (between 8,000 and 12,000 rev/min), is varied by an external controller on a belt and the pump delivers 3 – 8 L/min. There are no valves and the blood washes the pump bearings. Blood damage is said to be 'minimal' with some patients taking warfarin while others show no clot formation with only aspirin. Due to the small size of this pump, patients with a small body surface area can be treated, which implies that children may be future patients. (Kukuy, Oz, et al, 2001).

This pump is connected in parallel with the natural heart, which is still beating. This means the heart is not carrying the full blood circulation load and this assists in recovery.

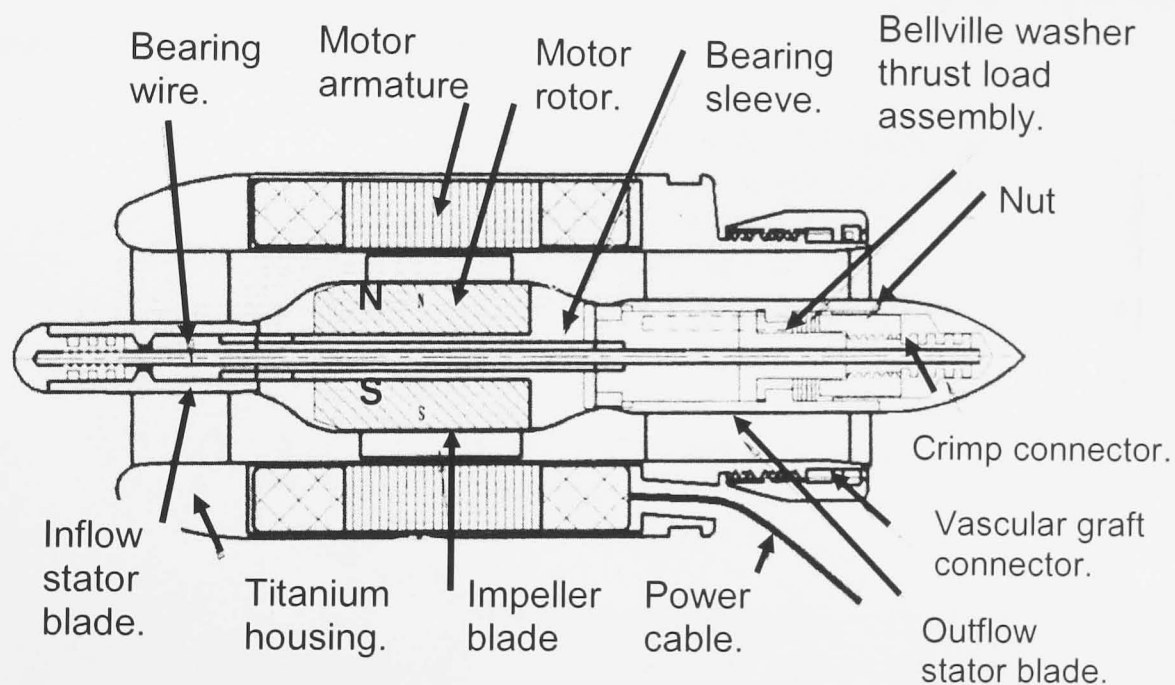


Cordes K, (2002). Project report, 2001-2002, p33.

Figure 2.28. Jarvik 2000 pump showing design features and the relative size of parts.

Figure 2.29 shows the parts and assembly of the pump. The main bearing of the pump is basically a length of stellite wire, diameter 0.9398 mm, running through the length of the pump and fastened at each end by crimp connectors in the inlet and outlet stators. A set of Bellville washers keeps the

wire under a tension load of 22.68 kg. with pyrolytic carbon sleeves providing both thrust and bearing surfaces. The major parts of the pump are titanium. The clearance between the tip of the rotor and casing is 0.1016 mm and wear on the bearings will be critical as the rotor may touch the inside of the casing. For this reason the bearings were changed to silicon carbide and a 5 months endurance test was conducted with the pump running at a speed of 10,000 rev/min. At the end of the test no measurable changes were seen for the shaft, the thrust bearing and at the inflow bearing at a measuring accuracy of 0.00254 mm. The outflow bearing showed wear of 0.00254 mm. It was estimated that the wear on the bearings will provide satisfactory duration for up to ten years. (Jarvik R, 1995).



Hetzer, R., Hennig, E., Loebe, M. (editors), (1997).

Figure 2.29, Assembly of the Jarvik 2000 blood pump.

The rotor has inbuilt magnets and is driven by a rotating magnetic field in the motor armature, set up by an external control system.

Three methods of supplying power to this pump have been developed.

The first uses the usual method of an electrical power cable that exits through the patient's abdomen.

The second method is a fully implanted pump and control system and uses the TETS system of energy transfer through the patient's skin to supply the pump with the necessary power.

The third method uses a plug and socket arrangement fitted to the patient's skull just behind one ear. From the fitting on the patients head, the power supply wires run under the bone of the skull, and then under the skin of the neck and body to the pump. This method reduces the infection risk relative to cables exiting through the abdomen. The external wires plug into the fitting on the skull and run down to a manual control and batteries on a belt or harness. Figure 2.30 outlines the installation.

A UK patient has lived for approx. 2 years with this pump implanted. The natural heart has improved, reinforcing that if it is 'rested' that there is a chance of recovery. There is nothing to suggest that the mainly non-pulsating flow has caused damage to the patient who appears have an almost normal life in terms of walking and exercising outside the hospital environment. (Westaby S. and Banning A., et al., 2000).

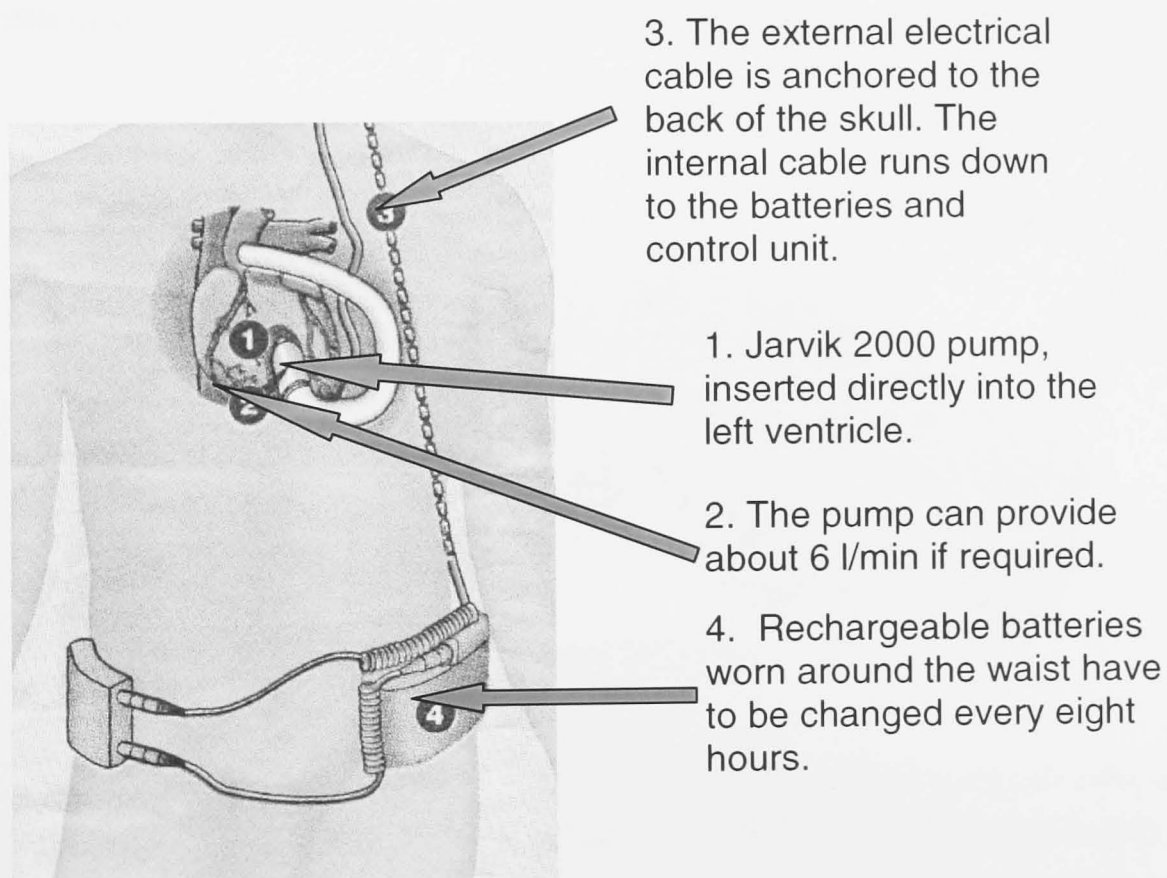


Figure 2.30. Installation of a Jarvik 2000 pump with the heart in situ.

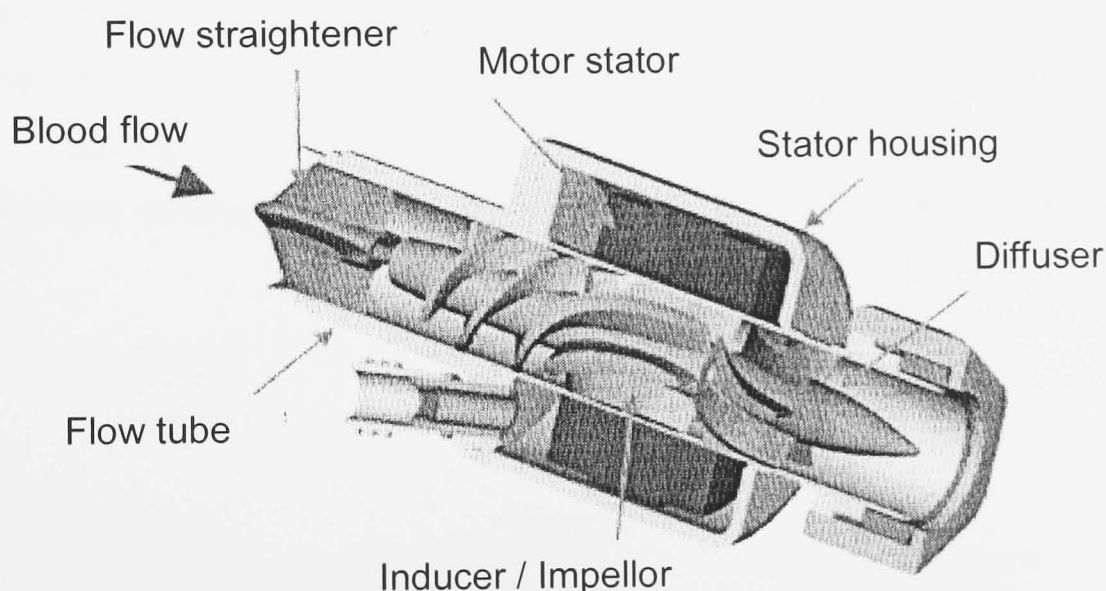
2.3.14 MicroMed-DeBakey pump.

An intraventricular pump similar to the Jarvik 2000 has been designed by the Baylor College of Medicine collaborating with the NASA Johnson Space Centre. Known as the MicroMed – DeBakey pump, it is approx. 31 mm

in diameter, 76mm long, (volume 57ml), has a mass of 0.095 kg and has a maximum output of 10 l/min. The pump takes blood from the base of the left ventricle and delivers into the aorta. In use, average pump flow is 3.9 to 5.4 l/min with warfarin anticoagulation treatment. With all intra-ventricular pumps, there must not be too much intake suction that will cause ventricular collapse or part of the ventricle to be drawn into the intake port.

As the pump delivers non pulsatile blood, it is possible to see a pulse on the pressure trace from the natural heart and the size of the pressure pulse is an indication of the condition of the natural heart. Some patients have shown an improved exercise tolerance and go home while awaiting transplantation. (Kukuy and Oz et al, 2001).

The improved condition is beneficial as recovery from transplant is quicker and with less complications if the patient is in 'good' condition before the operation. A sectioned pump is shown in Figure 2.31.



<http://www.micromedtech.com/productcomponent.htm>

Figure 2.31. Section through the MicroMed DeBakey pump.

It is noted from Figure 2.31 that the blood flow pattern changes as it passes through the pump. The first section of the impeller looks similar to an Archimedean screw where the flow will be axial. Passing into the centre section of the pump, the inducer / impellor appears to generate a rotary flow, imparting kinetic energy to the blood as the rotor spins. The diffuser appears to have blades to convert the rotary flow into axial flow before the blood

leaves the pump. This is probably generates pressure at the outlet of the pump. It is assumed that excessive stresses are not set up in the blood with the changes in flow direction through this pump.

2.3.15 HeartMate II pump.

This axial flow pump uses blood immersed ceramic and jewel pivot bearings for the rotor. There should be no reactions with blood from the bearings as they are biocompatible but the reaction of blood as a lubricant is not fully understood. (Chapter 2.3.4, the Vienna pump, discusses some aspects of blood lubricated bearings.) A cross section of the pump, Figure 2.32, shows a power cable leading out of the pump but a future version is planned using a fully implanted control system with TETS power transfer. There is no volume compensation with this pump which is fitted in the usual position between the left ventricle and the aorta. As this type of pump is 'small' compared to pulsating type pumps, it can be placed closer to the heart with shorter inlet and outlet tubing.

The pump is undergoing clinical trials and is expected to be suitable for adolescent, paediatric and adult patients. Schöb and Loree (Date accessed 2005).

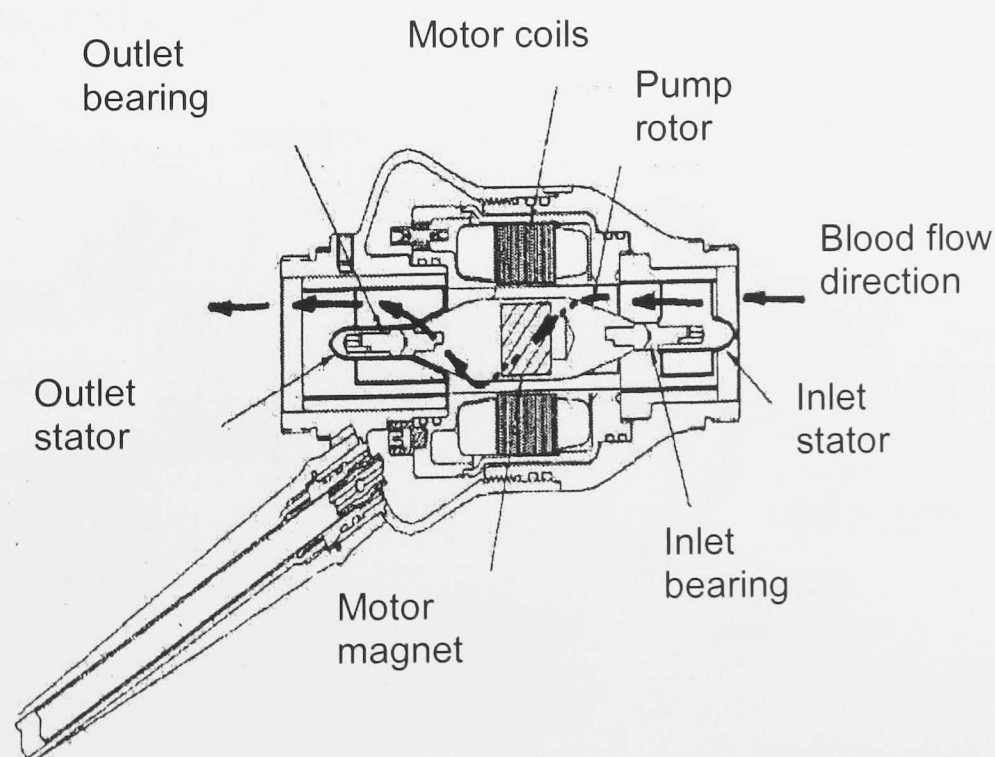
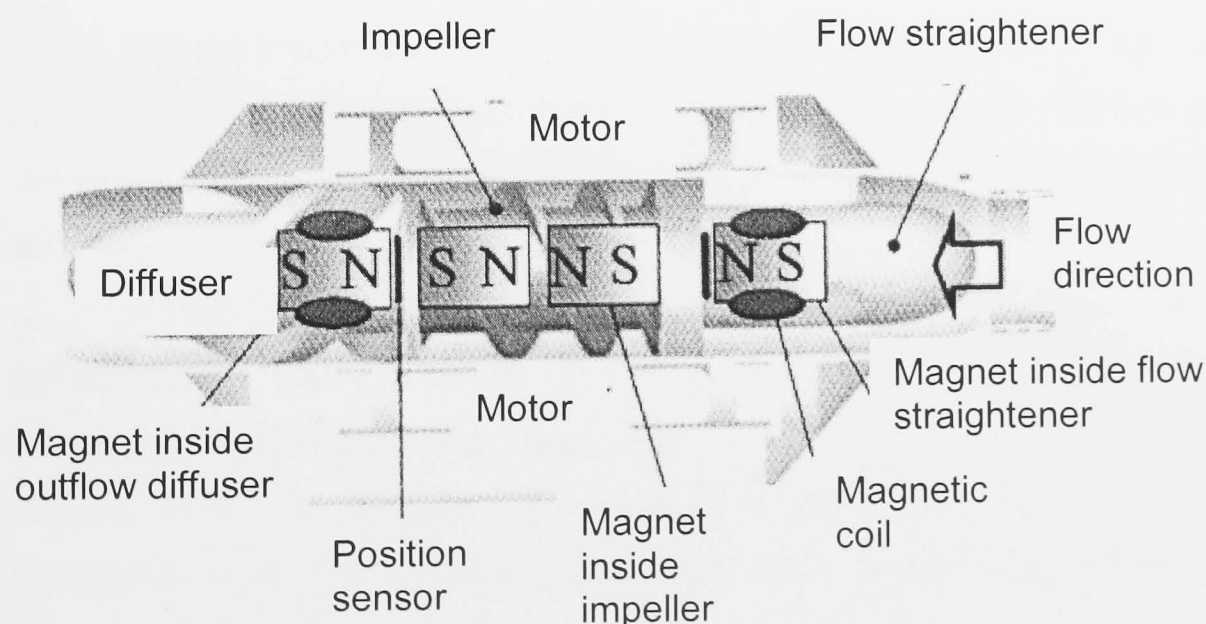


Figure 2.32, Cross section of HeartMate II pump showing general arrangement and blood flow direction.

2.3.16 InCor Berlin pump.

Huber C.H., Tozzi P., et al., (2004), describe an axial flow implantable blood pump with magnetic suspension to position the rotor inside the pump and a magnetic drive to pump the blood. The rotor does not have bearings to wear, and assuming no collisions of rotor and pump casing, the pump life is considered to be 'infinite'. This pump is considered to be a 'fourth' generation pump because it is 'contact free' in terms of bearings. The titanium pump, mass 0.2 kg, is 0.114 m long with an installed volume of 80.6 ml.

Figure 2.33 shows a section of the pump and it should be noted that the rotor is free to move axially inside the pump. From the position and speed of the rotor the differential pressure across the rotor can be calculated. Changes to the pressure difference cause the rotor to move axially inside the pump and if position sensors determine that the rotor is not located correctly inside the pump 'a counteracting force is generated in the magnetic suspension system' to return the rotor to its correct position. This statement is not fully understood but it is assumed that the rotor speed is changed in order to keep the rotor in the optimum position inside the pump housing. The amount of energy used to keep the rotor in position is used to calculate the flow through the pump.



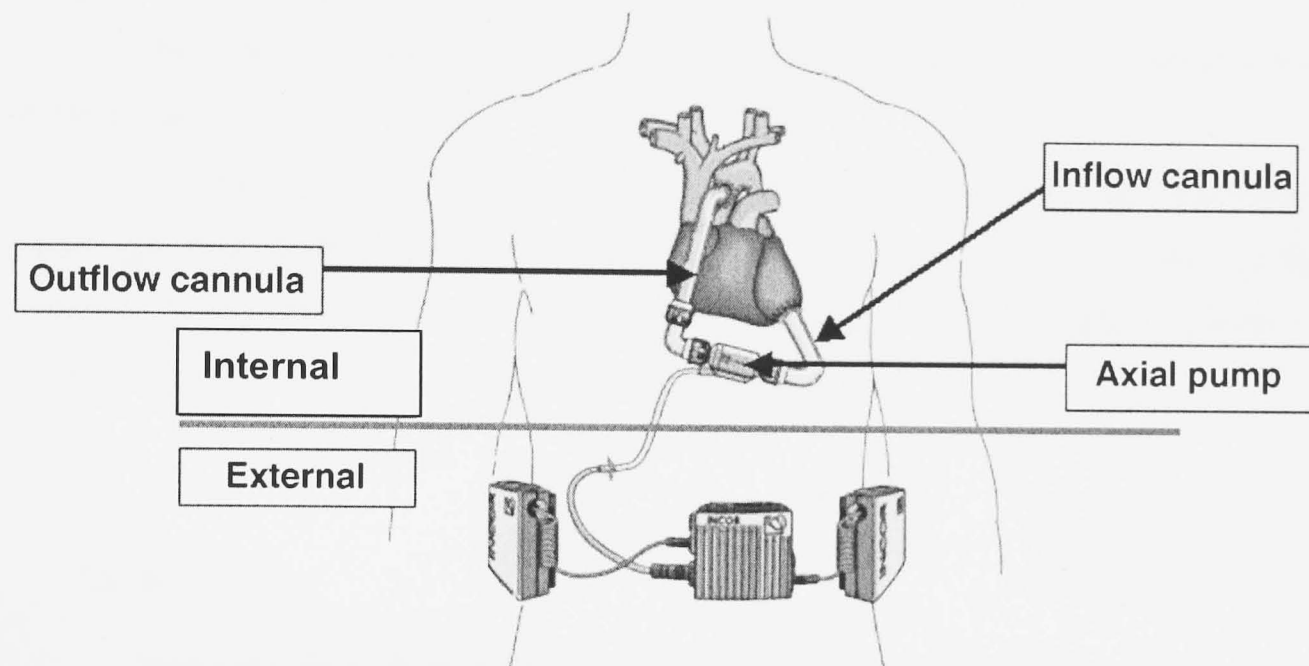
Huber C.H., Tozzi P., et al., (2004) and

<http://icvts.ctsnetjournals.org/content/vol3/issue2/images/large/4000362.336.gr1.jpeg>

Figure 2.33 General arrangement of the InCor Berlin axial flow pump.

At a speed of 5,000 to 10,000 rev/min., the pump delivers up to 5 l/min at a back pressure of 100 mm Hg. at a power consumption of 2 - 4 W.

The pump is implanted in the usual position between the left ventricle and the aorta. The power supply and control cables exit the body through the abdomen and a controller and batteries are carried on a belt or harness. Figure 2.34 shows the installation relative to the patient.



<http://icvts.ctsnetjournals.org/cgi/content/full/3/2/336>

Figure 2.34 showing the installation of the Incor Berlin blood pump.

Animal experiments over 6 hours gave an average flow of 2.5 l/min at a pressure of 58.8 mm Hg at 6590 rev/min. This is about half the quantity delivered by the human heart. At the end of the tests all pumps were free of thrombi.

A trial of 24 patients showed 'good' results in that 5 died, 2 recovered sufficiently to leave hospital and three received heart transplants. The remainder stayed on pump assistance. It was suspected that thrombi had formed with two patients as pump flow decreased but motor current increased. Treatment appeared to resolve the problem without removing the pump or surgical intervention.

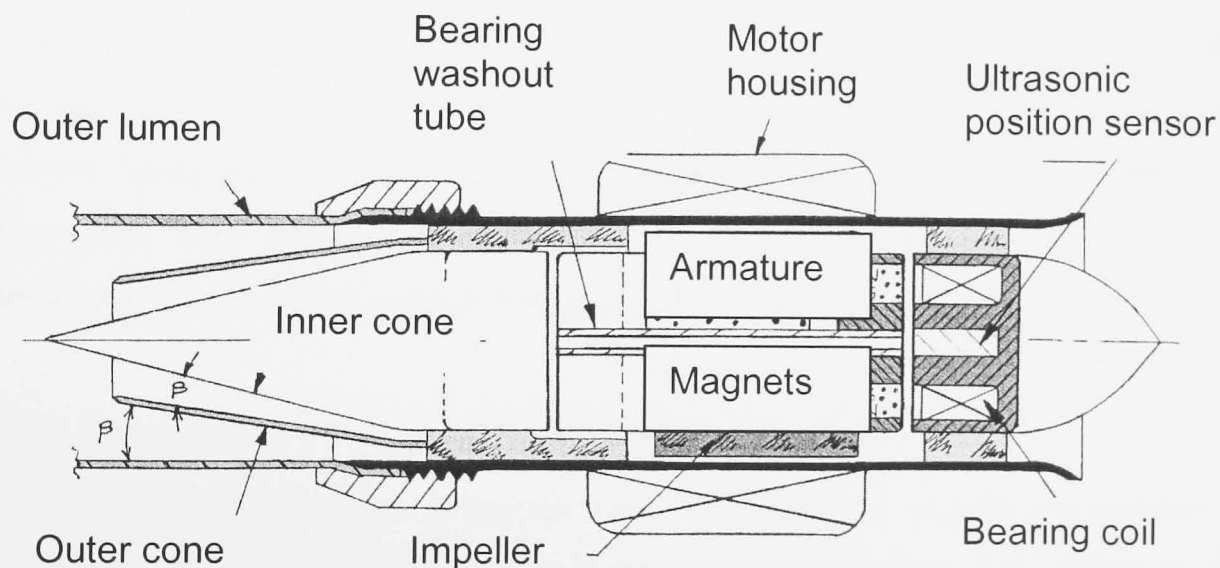
Goldstein D., (2003), reports that the blood contacting surfaces of the pump were bonded with heparin but the thrombus problem still occurred. This indicates problems either in materials or in blood flow characteristics.

2.3.17 Magnevad pump

Goldowsky M., (2004), describes another 'fourth' generation, axial flow, implantable pump that uses magnetic suspension to locate the rotor and a magnetic drive to pump the blood.

The pump is intended to deliver pulsatile flow by constantly varying the rotor speed.

The rotor passive magnetic suspension system does not have any bearings, and with no collisions, the life of the pump is considered to be 'infinite'. It has a mass of 0.082 kg, is 30 mm long and an installed volume of 25 ml. Figure 2.35 shows a section through the pump.



Goldowsky M., (2004).

Figure 2.35. Section through the Magnevad pump.

The pump is made of three parts, an inlet stator, a magnetically suspended impeller and an outlet stator with a double cone arrangement. The cone(s) are intended to smooth the blood flow to minimise blood turbulence and the generation of blood clots. The cone feature is unique to this pump.

The rotor is positioned by magnets at either end of the rotor and an ultrasonic sensor is used to determine the axial position of the rotor. If the

rotor is not in the correct position an automatic control system changes the rotor speed to keep the rotor in the correct axial position.

The small hole in the centre of the rotor is intended to allow a small amount of blood to wash the gaps between the ends of the rotor and the inlet and outlet stators.

The same basic design of pump, suitable scaled, is intended to be suitable for adults as well as children.

The maximum speed of the pump is 15,000 rev/min but will deliver 5.2 l/min at 13,000 rev/min at a pressure of 120 mm Hg. A speed of 11,000 rev/min generates a pressure of 80 mm Hg and cycling the pump between these two speeds is intended to represent the typical blood pressure variation for an adult.

For the pump to be running safely it will always have to be delivering some blood flow and this flow will have to be enough to prevent reverse flow through the pump and back into the heart. If the pump is running too fast on the pumping beat of the heart, it may generate a pressure behind the aortic valve that will not allow the valve to open. If the pump is running too slow when the heart pumps, the heart may generate enough pressure to open the aortic valve and attempt to push blood back through the pump. More information is needed to explain the operation of this and similar systems.

2.3.18 Summary of non-pulsatile flow pumps.

Pumps delivering non-pulsatile blood flow are the second generation of blood pumps but there does not appear to be full agreement amongst the medical profession that these pump types are fully satisfactory for the purpose intended because they do not give a pulsatile blood flow. Experience so far seems to show that the body adapts to this type of blood delivery method. Other researchers claim that pulsatile flow is necessary to maintain the blood flow through the brain and kidneys. More studies and data are required to answer this question.

There is evidence of less drug support required to keep the blood from clotting with non-pulsatile pumps. Kukuy et al. (2001) report that some

patients only require aspirin instead of warfarin as anticoagulation treatment, compared to pulsatile pump users.

There are no shaft seals to fail and no need for seal flushing systems with magnetically suspended systems. If this method of suspension can be guaranteed to be satisfactory, it would seem to be an ideal method. There could be problems with this method however concerning any interruption of power supply causing the rotor to contact the inside of the casing.

A listing of some of the artificial heart types and applications is shown on Appendix 4.

Control systems for rotary pumps appear to be simpler than with pulsating pumps, as patients can be sent home with portable, manual controls in use.

As with all other pumps, battery life is an important restriction to patient mobility. A battery with a life of 8 hours barely covers the overnight sleeping period and must restrict the day time activities if the patient needs to leave home.

The next part of the chapter will describe the power sources provided for blood pumps. It will show that power sources are provided in two main versions and that many power supply methods have not been successful.

2.4, Power sources.

This part of the chapter will describe the power sources that are under hospital control.

Brief descriptions of power sources that have been tried but have not been successful will also be discussed.

2.4.1 General description.

Those patients recovering following heart surgery may need a heart assisting device to recover satisfactorily. These devices are normally powered from the mains electricity supply through a control console and examples of would be the peristaltic roller pump, the intra aortic balloon pump, the Berlin pump and the BVS 5000 pumps described previously.

For patients who have received a blood pump, they can be supplied with mains or battery power depending on whether they are in or out of hospital.

With an 'in hospital' system, support is dependant on patient recovery and there is no restriction on the length of time power is supplied. With an 'out of hospital' system, batteries will need constant attention.

It is advantageous for a patient to be as fit as possible before the transplant and they are encouraged to exercise. For this they may use a battery powered pump so that they can be mobile which implies the system has to be light and easy to use.

The length of time the batteries last before recharging can be quite short, 6 to 8 hours is typical and several sets of batteries may be needed to ensure that one set is always ready.

The HeartMate, (Figure 2.5), and the Berlin heart, (Figure 2.8), are systems that can be portable.

With the Berlin heart 'Excor' system, the patient can leave the hospital if required but is tethered to a power system that is either carried around as a backpack or is pulled around in something approaching the size of a shopping trolley. While not the ideal way to live, the patient can exercise, keep reasonably fit and have some measure of independence while recovering or waiting for further treatment.

The system is designed for biventricular operation and uses two independent drive units. They are coupled so that if one unit fails the other continues to operate and supplies power to both blood pumps. The 12 volt batteries will give a service time of 6 hours and emergency batteries give 30 minutes of power. Indicators give information on battery charge condition and warnings of any problems with the equipment. (Drews et al, 2000).

The length of time some patients using artificial heart support can be short, for example hours to days. Other patients have to wait considerably longer. Loebe et al., (1995), report that The German Heart Institute, Berlin has one patient that has used artificial support for 500 days. Some patients at that hospital have been known to leave the hospital on shopping trips and other excursions.

For those patients that have been fitted with a non pulsatile pump, the size of the batteries and control system can be reduced as the system takes less power and it is easier to control this type of pump. With this system the batteries and control system can be carried on a belt or lightweight harness and it does not need a backpack or trolley.

The pump controls do not need to be automatic, the Jarvik 2000, uses a manual control for pump speed. The batteries last around 6 to 8 hours before recharge is needed. The power consumption is around 6 – 7 watts at a blood delivery of 5 – 6 l/min. (Westaby S., 2000).

The power consumption for an artificial heart depends on the type of device. Some of the pulsating artificial hearts, required 15 to 20 W and concern was expressed in terms of heat generation affecting the blood. The later, rotary, pumps require power ranging from 70 W for the Hemopump, (an intraventricular pump), to 6 W for the DeBakey pump, (an axial flow pump), with any heat generation being removed by the blood. A temperature rise of approximately one degree through the pump does not appear to affect the blood. This is not surprising as a fever temperature of 40° C, (104° F), is possible with full recovery.

Radial flow pumps have a similar power range from 4W to 25W. (Reul and Akdis, 2000). Specialised batteries have been developed for blood pump applications. Lead-acid batteries are unsuitable for implantation, reasons

include 'gassing off' during charge and discharge and the time taken to recharge.

The power supplied by a battery compared to the weight is important for a portable device. The power-to-weight ratio varies between various battery types and Table 2.9 is a comparison considering lead-acid gel batteries having a power-to-weight ratio of 1. This indicates that lithium batteries hold promise for implantable batteries at present.

<u>Type of battery</u>	<u>Power to weight ratio</u>
Lead- acid gel	1
Nickel Cadmium	1.165
Nickel – Metal Hydride	1.487
Lithium	2.539
Zinc- Air	4.261

Starnes T., (1996).

Table 2.9 Comparison of battery types and power to weight ratios

Rowles, et al., (1992), quote a combination of implanted nickel-cadmium rechargeable batteries that supply power for 20 minutes and external silver-zinc batteries which are enough to drive a blood pump for 5 hours.

Lithium batteries have been developed for consumer electronics, (mobile telephones, electric vehicles) and have been considered for implantable medical devices. For these applications the batteries are strongly built, corrosion resistant and with low magnetic characteristics. These are the ideal physical attributes for an implanted battery. Bruce, (1997), provides a technical explanation of the chemistry involved in the operation of these batteries. The paper indicates that each cell has the advantage of storing more than twice the energy of 'normal' rechargeable batteries of the same size and mass.

Batteries are made for implanted devices by the Saft Lithium Battery Division, Saft lithium battery, (2002). For example, the cells for the Berlin heart are flat, to giving a better fit to the body, and measure 17*48*65 mm. Each cell weighs 0.121kg, has a nominal voltage of 3.6V and capacity rating

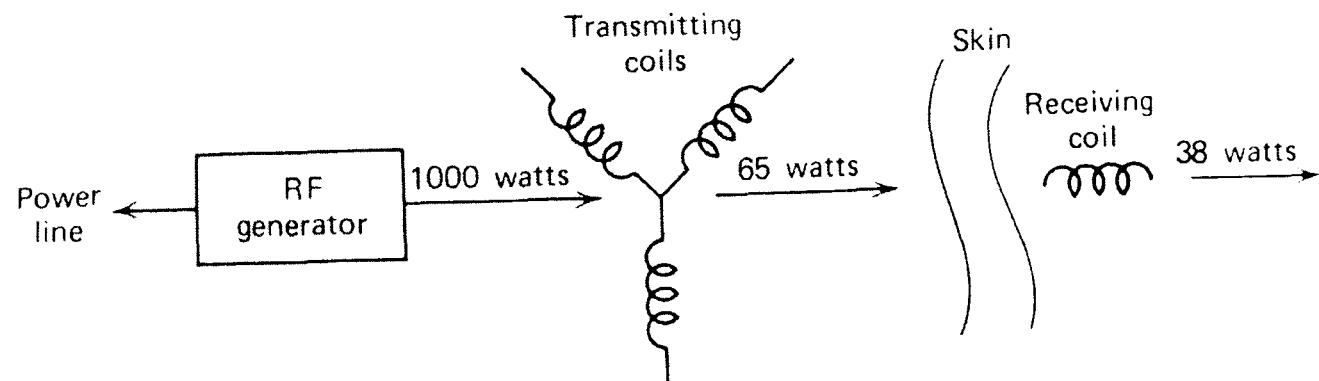
of 4.6 Ah. Stacking cells together gives the required voltage for the heart assist device. These cells can operate the pump for up to 10 hours without recharge and using specialised technology can recharge in 60 minutes. It is claimed that these batteries will recharge to greater than 85% of the 'new' capacity after 500 charging cycles. These figures would indicate a working life of about 7 months assuming the 85% recharge capacity.

A method to determine the useful life for the battery must be found to ensure safety for the user as there is no point in expecting 8 hours of use if the battery fails at 5 hours. A practice of replacing batteries after a given number of years may have to be implemented even though they still seem to be satisfactory. There was no information of the cost of the batteries.

2.4.2 Implanted power systems.

A power system that has potential is the TETS system, (Transmission of Energy Through the Skin). This has been developed to enable implanted pumps to be powered from an internal battery and control system. The internal batteries, which last about half an hour, are recharged through the skin, which eliminates the prospect of infections.

The first system used two pancake coils to transmit and receive the energy. One coil was placed outside the body, the other was implanted. The system was stated to have transmitted 38 watts with an efficiency of 95% using a frequency of 465 kHz. Figure 2.36 shows the circuit diagram of this system.



Myers and Parsonnet, (1969).

Figure 2.36 Electrical circuit diagram of a TETS system.

It was important that the two coils were in close proximity in order to transmit the required energy and the system has been tested for this using an animal. The results indicated enough energy was transmitted as long as the receiving coil was within 1m³ of the transmitting coil. (Schuder and Stephenson (1963) and Schuder, Stephenson and Townsend, (1969). Other systems have been developed that run from battery power.

A similar system by Myers and Reed, (1968), used coils strapped externally to the body. The electrical coupling of the coils was set up so that any slight misalignment of the coils would not seriously affect the amount of power transmitted. The power and efficiency stayed reasonably constant for an air gap spacing of up to 17mm and an 'out of alignment' of the coils of up to 10 mm. The system used a lower frequency of 13 kHz and achieved an efficiency of 91%. The system was stated to be capable of charging the internal batteries without noticeable skin damage. The patient can be trained to recharge the internal battery using a harness to position the external coil. The package is self-contained and does not need any external controls. Figures 2.37 and 2.38 show the system proposed.

A harness carries the external batteries and control monitor. The external coil placed adjacent to the internal coil transmits the power through the skin required to run the internally implanted blood pump and control system.

External electronics package.

Internal electronics package.

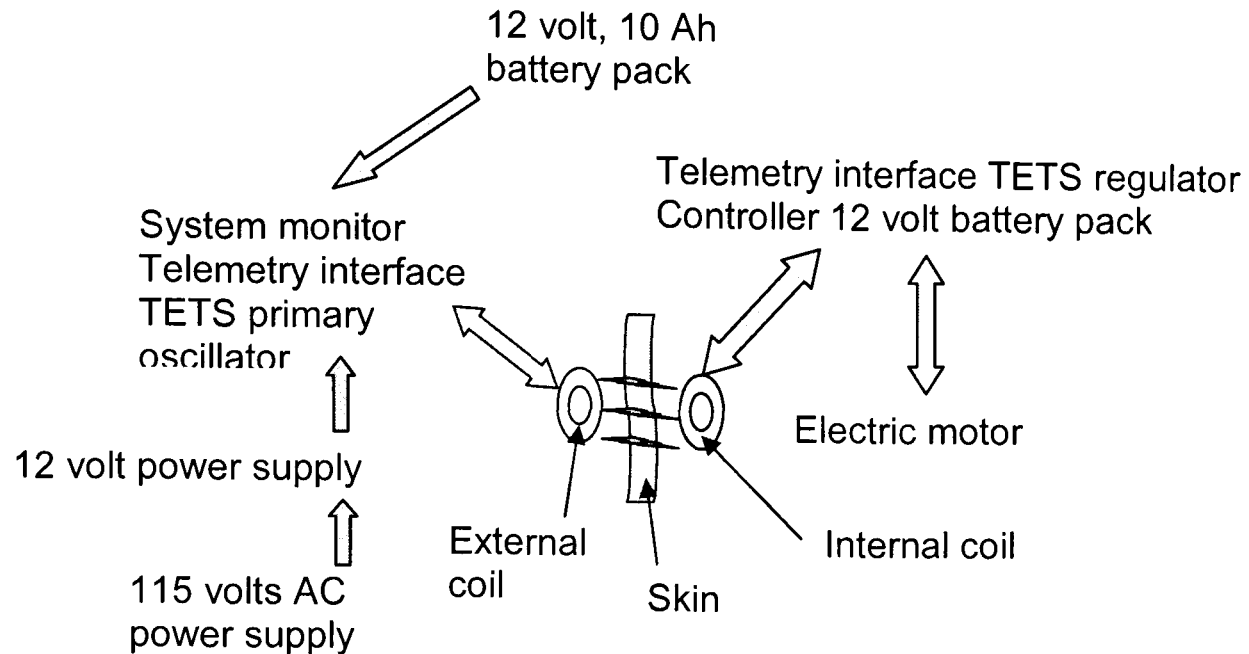
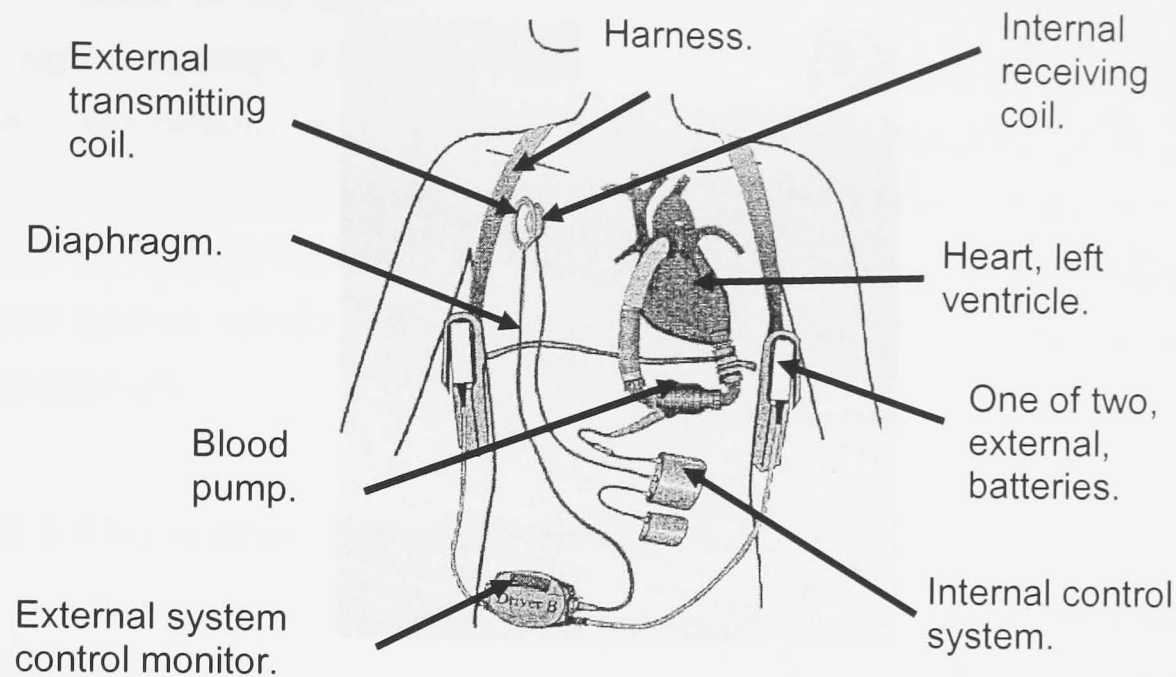


Figure 2.37. Schematic layout of a TETS system,
Transmission of Energy Through the Skin.

U.S. patent no. 5,109,843.
From Engineering in the Heart and Blood Vessels,
Myers and Parsonnet, 1969.

Another TETS system is proposed by Goto, et al., (2001). This system uses near-infrared light passing through the skin to charge an implanted battery. It is claimed that 17 m of charge time can send enough energy to run an implanted pacemaker for 24h. The rise in skin temperature was stated as 1.4°C during charging. Assuming a 'nominal' temperature of 37°C this would give a skin temperature of 38.4°C. This temperature indicates that the patient would feel some sensation while the system was charging. This system is not suitable for a blood pump as a pacemaker uses a much smaller amount of power than an implanted blood pump, (mW compared to W), and with this particular system the increased length of charging time necessary might not be enough to power a blood pump.



Gagnon D. et al, (2000), *Perfusion* (2000); 15: pp345-354.

Figure 2.38. Arrangement of TETS system relative to patient.

Starner and Maguire, (1999), claim that a temperature of 43 °C for extended periods can cause the skin to burn although temporary contact at higher temperatures, (48 °C for 10 minutes), can be maintained without damage.

A 'safe' skin temperature of 41.5 °C was selected by Lele, (1987), following a review of medical literature. The effect of using the same area of skin repeatedly over an extended period of time is unknown and it may be necessary to move the TETS charging point to reduce damage. This movement will not be possible with the implanted coil although enough power may be transmitted at a lower efficiency if the two coils are offset relative to each other and using a different area of skin each time.

Another development to transmit power through the skin is offered by Matsuki et al., (1998). They propose a flexible implantable transformer, 1 mm thick, made from amorphous magnetic fibres and with a low heat profile. A transformer area of 2,100 mm² was stated to be able to supply 6 W with a negligible temperature rise. (This implies an implanted device around 46 mm square). If the external part of the transformer was the same size, or larger to allow for misalignment, this could be included as part of clothing worn next to the skin.

Ozeki et al., (2003), proposed a method of supplying TETS power through a specially built floor combined with special shoes. They indicated that it was possible to meet the power requirements with a DC to DC power efficiency greater than 60% as the patient moved around on the floor. This would be ideal for a patient living in one room in the home but not elsewhere. There was no mention of effects on electrical equipment in a house and no cost estimate.

2.4.3 Alternative power supplies.

Myers and Parsonnet, (1969), suggested generating electricity by placing two electrodes of differing metals into the body. This is technically possible but the corrosion products from the electrodes are likely to be toxic and the amount of electricity produced may not be enough to run a pacemaker, which requires 10 to 20 mW.

Myers and Parsonnet, (1969), also suggested using thermocouples. The problem with this method is the lack of a cold junction because the human body regulates temperature to a very narrow band. The solution proposed creating an artificial hot junction with the body acting as the cold junction. This used a radioisotope powered device with Plutonium 238 as a power source. This would have a half life of about 90 years and a running temperature of 300 °F. This would give a temperature difference of about 200 °F but the power generated was not considered enough to run an artificial heart device.

Talaat, (1967), proposed implanting two identical, non-corrodible electrodes into the heart to make a cell and generate around 100µW. This is enough to run a pacemaker but not enough for a blood pump.

Natural movements of the body have also been suggested as a power source. Starner, (1996), suggested a small turbine generator implanted in the blood stream and piezoelectric generators operated by the movement of the heart.

Epstein, (2006), is attempting to build a 10 watt gas turbine engine on a microchip. This amount of power will run some of the smaller blood pumps.

Another alternative power supply that has been tried to assist the failing heart is skeletal muscle augmentation.

Pepper, (2000) and Salmons and Jarvis, (1990), proposed partially removing the selected muscle from the patient and grafting it to the failing heart. A pacemaker device then stimulates the muscle to contract at the correct speed and time effectively squeezing the heart to maintain blood flow.

Pepper, (2000), also proposed wrapping a conditioned muscle, around the aorta to provide additional pulsation effects.

The results of these types of procedures are not particularly good and much work remains to be done before this procedure becomes a realistic option for patients with heart problems.

These methods have identical problems to the biological –mechanical pump system, described previously in this thesis, of muscle conditioning. They have not achieved acceptance.

2.4.4 Pump control systems.

The only method of controlling the output of a pump appears to be by changing either the speed of rotation for a rotary pump or the 'beat' rate for a pulsating pump.

Catanese, et al., (1996), indicate that the *"HeartMate electric pump is usually operated in the 'automatic' mode, so that the pump ejects when it is 90% filled. In this way the flow increases when patient activity increases and decreases when the patient activity decreases"*. From this it is understood that the pump fills passively and a proximity sensor is used to determine when the pump is filled to a predetermined amount. When that condition is reached, power is applied to empty the pump.

Youngson, (1993) suggest automatic control could be based on the pulse rate of the existing biological heart. This requires the heart to be in place and functioning.

Power adjustments according to flow are proposed with the Incor Berlin pump described previously in Chapter 2.3.16. This uses a position sensor to adjust the pump speed and flow.

Karl, (1997) suggests using a flow measurement device to monitor the blood flow rate and adjust the pump speed accordingly. This method will have problems in terms of calibration and stability.

Rosenburg, (1995), indicated that it was preferable to use few or no sensors in the control system as they can suffer from signal 'drift' over time which makes them extremely difficult to deal with *in vivo*.

The pumping rate can be changed manually by the user using an external control. This is the method used with the Jarvik 2000 pump but how the patient knows which setting to use is not stated.

2.4.5 Summary.

Methods of powering blood pumps have been described. For the patient in a hospital bed, power is no problem. For a patient outside the hospital, they must always be conscious that life depends on batteries and in this sense 'life' is only 6 – 8 hours long.

Improvements to blood pumps have reduced the power consumption from around 70 W to as little as 4 W. Improvements to control systems, by making them smaller and lighter, have allowed patients to be mobile. Both these together have allowed some patients to go home and live a relatively normal life except for the problem of power supply.

Improvements to batteries have improved the electrical storage capacity, the power to weight ratio and the amount of time required to recharge, but further improvements are necessary.

All batteries have a 'useful' life and there comes a point where they will not last the expected length of time and it can be considered to be 'worn out'. There is no mention of methods to determine how much life the battery has left. The obvious method would be to replace the battery at 'timed' intervals. This could be months or years or the number of times the battery has been charged and discharged. This may be expensive but would be simple to do. For an external battery this will be no problem but for an internally implanted battery this will mean the patient undergoing another operation, to replace it, with all the attendant risks.

The development of the TETS system shows promise as it removes the problem of infection caused through pipes or cables passing through the skin. Heating of the skin while charging the internal battery has been noted. This does not appear to be a serious problem as long as the temperature rise is less than 5°C.

There does not seem to be alternative methods to supply power to run an implanted blood pump of the present generations without the use of batteries. Attempts have been made to use the body itself but these have only produced enough power to drive pacemakers.

Control systems need to be as simple as possible. Manual control is possible as used on the Jarvik 2000. Proximity sensors can be used as a signal generating device to control the pulse rate of a pulsating pump or the speed of a rotary pump. There are likely to be problems with calibration and 'drift' with implanted sensors. These cannot be corrected without a surgical procedure and the attendant risks to the patient.

The next part of this chapter will describe the performances obtained from various pump types compared to the human heart. It will outline the delivery and pressures obtained and the pump applications. Comparisons will be made of the relative sizes of each pump and the outputs delivered.

2.5 Comparison of performance.

It is generally accepted that the human heart pumps blood at a rate of about 5.5 l/min at rest under normal circumstances, Peterzén B., et al., (1996) and Dawson T.H., (1991). Any manufactured device to replace the function of the heart should therefore be capable of pumping a similar amount.

It is noted that there does not seem to be a recognised specification for the performance of a pump to replace or assist the human heart. This implies that manufacturers could be working to different specifications and therefore makes it extremely difficult for surgeons and others to compare what is offered. This omission could be of great importance to the condition of the patient.

Table 2.10 shows the quoted delivery, the installed volume and the mass of the pumps listed, together with power consumption from the available

literature. These quantities are expressed as a ratio compared to the human heart as far as possible.

The estimated figures for the human heart are taken as:-

Volume pumped 5.5 l/min, installed volume 622 ml, mass 0.283 kg. Grays Anatomy, (1995). (The installed volume is probably over estimated as it is calculated as a rectangular solid and takes no account of the shape of the heart). The Physics Factbook quotes the power consumption of the heart as 1.868 W.

Pump types are denoted as, P = pulsating flow, R = rotary flow, A = axial flow.

Bearings = rotor is mounted on blood lubricated plain bearings.

Mag. lev = rotor is magnetically levitated.

Some of the pumps listed are capable of pumping more than the accepted delivery of the heart of 5.5 l/min. The pumps that deliver less than 5.5 l/min may be useful for children as they are not suitable for adults needing full support.

Not many references quote the installed volume of each pump. The human body cannot have spare room inside it and if a pump is installed organs may have to be moved aside or compressed to accommodate the pump. What damage or discomfort is this likely to cause?

There is a trend in the mass of each pump type. The pulsating pumps can be heavier than the human heart, (2 to 3 times), while the rotary pumps can be lighter than the human heart by up to 0.85 the mass. The mass of the axial flow pumps can be less at around 0.34 although the HeartMate II appears to be heavier than the human heart.

The mass of the pump is important is the sense of support inside the body. This topic is covered in greater detail in Recommended Features, Size and Installation, Chapter 2.6.3

<u>Pump type / name</u>	<u>Volume pumped</u>	<u>Installed volume</u>	<u>Mass</u>	<u>Pump type</u>	<u>Power consumption</u>
Human heart	1	1	1	P	1
Abiomed 5000	1.09	N A	N A	P	133.8
HeartMate 1	1.96	0.63	2.4	P	N A
Berlin	<1	Various but <1	<1	P	N A
AbioCor (TAH)	1.8	N A	3.21	P	N A
Baylor (TAH)	1	0.82	2.19	P	N A
Novacor	1.55	N A	N A	P	N A
Spiral vortex	0.55	N A	N A	P	N A
Radioisotope pump	10.73	1.96	5.3	P	N A
Pendulum	N A	N A	N A	P	N A
Aachen	N A	N A	N A	P	N A
Vibrating flow pump I	0.36	N A	0.18	P	N A
Vibrating flow pump II	0.01	N A	N A	P	N A
Baylor rotary	1.55	N A	N A	R + bearings	N A
Vienna, ball bearings	1.8	N A	0.85	R + bearings	N A
Vienna, single pivot	1.27	N A	N A	R + bearings	N A
AB 180	0.55	N A	N A	R + bearings	N A
Venrta Assist	1.8	0.2	1.05	R + mag. lev	N A
HeartMate III	1.27	0.18	0.12	R + mag. lev	4.34
Intraventricular	N A	N A	N A	A + bearings	N A
Hemopump	N A	N A	N A	A + bearings	37.47
Jarvik 2000	1	0.04	0.32	A + bearings	1.61 – 3.75
MicroMed DeBakey	1.8	0.09	0.34	A + bearings	3.21
HeartMate II	1.82	0.2	1.2	A + bearings	N A
InCor Berlin	0.91	0.13	0.71	A + mag. lev	1.07 – 2.14
Magnevad	1.8	0.04	0.29	A + mag. lev	N A

Table 2.10. Comparison of the various types of blood pumps.

2.5.1 Summary.

Of the pump types listed, not all have full information on volume pumped, installed volume and mass. The comparison shows that most pumps can equate or exceed the pumped volume of the heart and therefore are intended for adult use and not children. An adult size pump is taken as a pump that will deliver at least 5.5 l/min.

The rotary and axial flow types have a reduced installed volume and lower installed mass than the pulsating type pumps.

Only 2 of the pumps are listed as Total Artificial Hearts, (TAH). Use of this type of pump seems to have declined and effort has been put into the development of heart assisting pumps rather than heart replacement pumps. This reflects patient recovery by heart assistance rather than heart replacement by an artificial pump.

The next part of the chapter covers the recommended features of any blood pump. It will cover the quantity to be pumped and the blood damage potential which is an important part of the performance of any pump.

2.6 Recommended features of a blood pump.

2.6.1 Quantity.

Any implanted pump must be capable of supplying enough blood to satisfy the needs of the patient. For an adult, requiring the maximum assistance, the amount of blood will have to be around 5.5 l/min. For a child this amount will need to be less in proportion to the size of the child. Some pumps have been built to satisfy child applications.

The pump will have to supply against the back pressure of the blood already circulating in the patient, (the diastolic pressure). For an adult this appears to be approximately 100 mm Hg but will vary depending on the condition of the patient. Hegyi T., Carbone M.T. et al., (1994), indicate that for premature babies, the diastolic blood pressure can be as low as 20 to 25 mm Hg, but the actual value will again depend on the medical condition of the baby.

The pump must be able to respond to any changes in performance with little or no time delay. For a pulsating pump this will mean rapidly changing

the 'beat' rate and a rotating pump will need to have the ability to change speed quickly. This indicates a pump with low inertia in the moving parts. A pulsating pump will be 'power hungry' as the mass of the moving part will have to be accelerated and decelerated on every 'beat' but with a rotary pump, once the rotating mass has been accelerated to the new speed, very little extra power may be required to keep it running. This will be an advantage in terms of battery life.

2.6.2 Power supplies and control.

There is a need for a satisfactory method of supplying power to the pump. Early methods using flexible mechanical drives have proved to be unsuitable.

There are two methods of operating blood pumps, pneumatic and electrical. An electrical power system to supply power to the pump is the TETS system where energy is transmitted through the skin to charge an internal battery. This prevents infection as there are no pipes/cables passing through the abdomen.

Changing the output of a blood pump is achieved by changing the beat rate or rotation speed depending on the pump type. Manual control can be used. Automatic control is possible with inbuilt sensors providing the control signals.

The batteries powering the pump must last a reasonable length of time. At present up to 8 hours is possible. A longer time would be preferred together with a shorter recharge time.

The pump must not cause any inconvenience as far as possible. This means little or no noise, vibration, heat or other discomfort.

2.6.3 Size and installation.

The physical size of the pump is important as there is not likely to be a large amount of spare room inside the body. Patients may be advised to eat many small meals often rather than one large meal as some pumps, typically the pulsating types, appear to be positioned close to the stomach. There does not appear to be any physical support for the pump and in such close

proximity, movement of the heart may touch the pump. Any pump movement may cause problems with strain on the pipes and anything the pipes are attached to. It is assumed that the pumps are supported in order to prevent any problems of this nature but there is nothing in the literature describing this part of the pump installation.

The equipment associated with the pump must not be too heavy to carry. The method mentioned in Chapter 2.4.1 using a backpack or trolley mounted system although acceptable to maintain life would not be acceptable socially and improvements must be found. The pump control system must be self contained and not require frequent adjustment or calibration. There must be alarms or indicators to show if the system is running properly or has a fault and which fault. The system must be 'fail safe' at all times.

2.6.4 Effect on the patient.

Although work has been done on the clinical aspects there seems to be little emphasis on the quality of life for the patient after implantation. This needs to cover characteristics from the pure medical condition to the mental outlook of the patient.

It is important the patient has the ability to continue to contribute to society and to experience their own social well being.

Quality of life conditions comprise: -

1) The realisation that life now depends on a man made device and that this device may have to last for the rest of life if a transplant is not possible.

What effect does this have on the patient? Satisfactory explanations of the 'after operation' conditions must ensure that the patient is aware of the implications.

2) Are there physical complications likely with this device such as bleeding, difficulty in handling the operation of the device, changing batteries? Is the patient competent to handle these crises, either with help or by themselves?

Is the patient competent to handle the drug regime necessary after implantation?

This will depend on the patients themselves although an amount of training and support will be necessary for success.

3) Are there any emotional problems, perhaps distress and anxiety, a feeling of being “not normal”?

2.6.5 Summary.

This chapter has described some of the features that are necessary for a satisfactory blood pump. The pump must be able to deliver the correct quantity of blood which is normally taken as 5 l/min for an adult. Lower deliveries will be necessary for children, the amount depending on the age and size of the child.

The pump must be able to supply this quantity of blood for an almost indefinite time period without causing any damage to the blood and without any medical problems to the patient. It is assumed that a small amount of blood damage accumulating for an extended length of time can be just as damaging to the patient as a large amount of damage over a short time. From the results of Leverett and Hellums, a maximum blood stress level of 100 N/m² may be possible for an indefinite time period but this is not proven here. The critical amount of damage is difficult to determine but from the previous results it is likely to be at a stress less than 25 N/m².

Stresses below this limit appear to be satisfactory, but additional evidence is necessary to prove the case for an indefinite time period. In this concept an indefinite time period is taken to be at least 10 years. Limiting blood stress levels are difficult to determine as they appear to be a combination of many factors. These include surface interactions, surface finishes, hydraulic forces, heating and exposure time. From these results it does not seem possible to determine with great accuracy the critical amount of blood damage before problems are likely to occur. A range of stress and time limits may be more appropriate, the limits of which, when applied to the individual patient may in some part depend on the genetic make up of the patient, the general health at the time and on age.

During this time period and with the pump running *in vivo*, some blood damage below critical levels may occur that are due to the pump. A healthy patient may have the capacity to withstand and recover from an amount of

blood damage, but the same amount of damage caused by the blood pump to a sick patient may be significant.

Of the three types of pump available, (axial flow, radial flow and pulsating types), the axial flow pumps have a shorter residence time than the pulsating pump but have appreciably higher stresses. The centrifugal pump has lower stresses than the axial pump but has longer residence times leading to the possibility of increased surface effects.

The development of rotary and axial flow pumps with magnetically suspended rotors has promise for a pump with a longer life than those at present.

The physical size of the pump needs to be taken into account as a pump that is 'large' compared to the available room inside the patient will not be suitable for use. Internal movements of the body organs must not be allowed to move the pump excessively and cause strain on anything that the pump is attached to. The connections to the heart and aorta must not be compromised in this respect.

Power supplies and the means of getting the power to the pump need consideration. The maximum 'life' for a battery before recharge is required is around 8 hours and improved battery technology is required to extend this time. Methods are being developed that enable power transfer without breaking the skin. This will reduce the possibility of infections entering the patient but care is necessary with skin temperatures while the energy transfer is taking place.

Some work has been done on the mental effects of installing a blood pump to the patient. The realisation that life is dependant on a man made machine for life has caused mental problems with some patients. Little work appears to have been done to investigate or understand these problems.

The next chapter will describe the materials used to build blood contacting devices with the advantages and disadvantages of each type in terms of manufacturing and properties.

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Chapter 3. Materials.

3.1 Biomaterials. General description.

For the purpose of relieving or solving an adverse medical condition, man made devices can be placed inside the human body. These are known generally known as implants. These are not new devices and the Table 3.1 shows a brief history and some milestone developments.

Any material used to make an implant is known as a biomaterial and a definition is given is given by Boretos and Eden, (Eds), (1984) where:-

'A biomaterial is any substance, (other than a drug), or any combination of substances, either synthetic or natural in origin, which can be used for any period of time, as whole or part of a system which treats, augments, or replaces any tissue, organ, or function of the body'.

This definition implies that the biomaterial interfaces with living tissue for a significant length of time. Bone plates and artificial heart valves are examples of implants in contact with blood or tissue.

Any biomaterial is always a 'foreign' material and must not have an uncontrollable adverse reaction when implanted in the host tissue, Parks J, Lakes, R, (1992). Any implant can start a reaction and care has to be taken to ensure the biocompatibility of materials and the construction and operation of the implant .

Year	Investigators	Development
Late 18th / 19th C		Various metal devices to fix bone fractures; wires and pins from Fe, Au, Ag and Pt
1860 - 1870	J. Lister	Aseptic surgical techniques.
1886	H.Hansmann	Ni plated steel bone fracture plate
1893 – 1912	W. A. Lane	Steel screws and plates, (Lane fracture plate)
1912	W. D. Sherman	Vanadium steel plates, first developed for medical use; less stress concentration and better corrosion resistance. (Sherman Plate).
1924	A. A. Zierold	Introduced Stellites, (CoCrMo alloy)
1926	M. Z. Lange	Introduced 18-8sMo stainless steel, an improvement on 18-8 stainless steel
1926	E. W. Hey-Goves	Used carpenter's screw for femoral neck fracture
1931	M. N. Sith-Peterson	First femoral neck fracture fixation device made of stainless steel
1936	C. Venable, W. Stuck	Introduced Vitallium, (19-9 stainless steel), later changed to CoCr alloys
1938	P. Wiles	First total hip replacement prothesis
1939	J. Burch, H. Carney	Introduced Tantalum (Ta)
1946	J. Judet, R. Judet	First biomechanically designed femoral head replacement prothesis,
		first plastics, (PMMA), used in joint replacements
1940's	Dorzee, Francechetti	First used acrylics, (PMMA), for corneal replacement.
1947	J. Cotton	Introduced Ti and its alloys
1952	Voorhees, Jaretzta and Blackmore	First successful blood vessel replacement made of cloth for tissue ingrowth
1958	Furman, Robinson	First successful direct heart stimulation
1958	J. Charnley	First use of acrylic bone cement in total hip replacement on the advice of Dr. D. Smith
1960	Starr, Edwards	First commercial heart valves
1970's	W.J. Kolff	Total heart replacement

Table 3.1. A brief history of implants

There are three reactions to an implant and these are described as tissue integrated reactions, rejection and thrombosis.

Tissue integration reactions can include the implant becoming 'walled off', when tissue grows around the implant. If the implant has to be removed this can cause difficulties for the surgeon. The integration reaction can be used advantage by allowing tissue to grow 'into' the surface of the implant to provide a blood compatible surface or a secure fixing for bone replacement.

Rejection of the implant makes it useless for the intended purpose and endangers the patient. An example of this is the transplantation of biological material from one person to another. The implant has to be carefully matched to the recipient for success. It is possible to suppress the rejection process with drugs but this reduces the ability of the patient to fight any infections and implies the patient may have to remain on anti rejection drugs for the rest of his or her life.

Implants in contact with blood can cause the formation of blood clots, (thrombi). Under these conditions the implant may not operate properly and the blood clots formed will be a danger to the patient.

Other problems include the possibility that the implant is a carcinogen which is a substance that produces cancer by means of direct contact with tissue, or by ingestion, or by inhalation. The contact with interior tissue is of particular concern as the cancer may not be easy to see and may have made considerable progress before it is discovered. Once discovered it may be extremely difficult or impossible to remove without severe problems to the patient. The implant must be free from infections. The implantation of a 'foreign' body can create large infection problems as the implant provides a region that the body's fighting cells cannot always reach.

Great care has to be taken with the choice of materials as there are very few that do not have some reaction at contact with blood. The incorrect choice of materials has caused problems. Zdrahala and Zdrahala, (1999) report life threatening events such as strokes have occurred. These events are not always fully understood but Fasol and Zilla, et al., (1987), indicate that material 'failure' may not always be due to the material itself but the failure of blood cells to successfully attach to the surface of a material. Their tests indicated that it would take up to 6 – 8 weeks to establish a satisfactory blood

compatible surface assuming that the cells were shear stress resistant. Following these results they concluded that *“the endothelialization of artificial hearts is not possible when such hearts are used for urgent ‘bridging’ before cardiac transplantation”*.

It has been suggested by Basmadjin, Sefton et al, (1997), that a truly inert biomaterial is not possible and that even trace amounts of factor XII, (one of the factors responsible for blood clotting), will lead to an initiation of the coagulation process

3.1.1 Reactions to implants.

The basic design of an implant is important. A pump with a large internal volume pumping a small amount of blood may not give a good internal ‘washout’ and stagnant blood has more opportunity to cause more problems than flowing blood. With a low blood flow rate, the longer the time the blood is inside the implant, the greater the risk of clot formation with reactions with blood contacting surfaces.

The suitability of a biomaterial is measured not only in the material characteristics but in coatings and surface finishes that provide resistance to the conditions inside the body. The body contains a degree of self-repair following the natural process of cell replacement but any artificial part does not have this characteristic and at best can only stay in the condition at implantation.

Coatings can be applied to components to prevent blood clotting, but these can wear away and lose effectiveness.

The surface finish of the implant is important and two approaches used are a smooth surface and a textured surface.

A smooth surface can be formed on metallic parts by electropolishing, a process related to electrochemical machining and is described in DeGarmo, Black et al. (1990.)

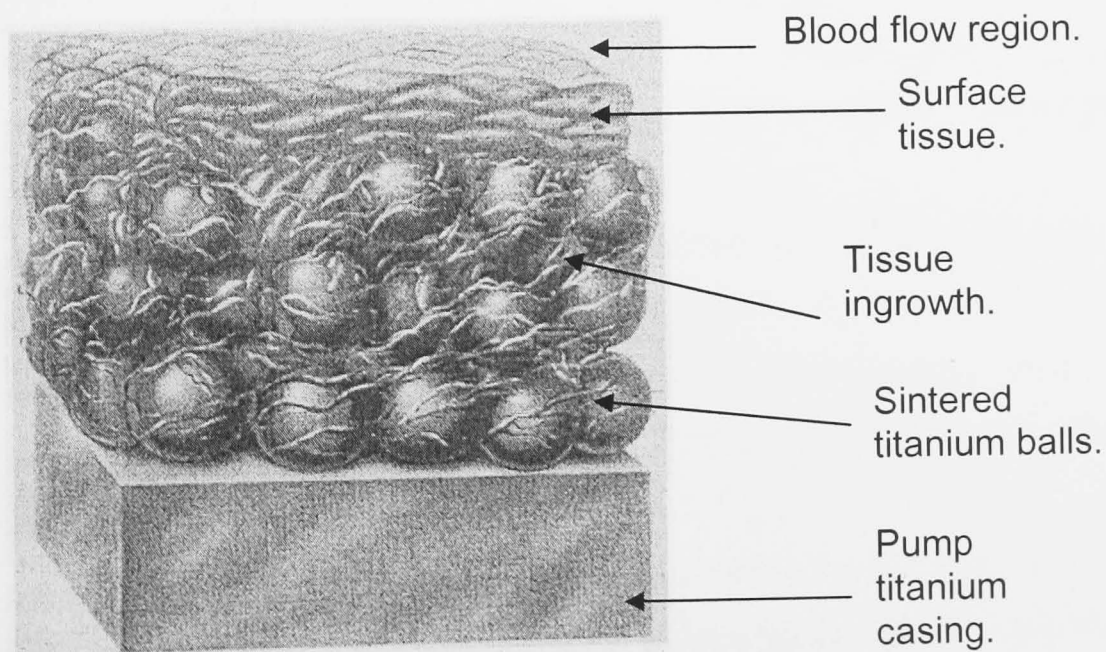
Dip moulding techniques can be used to manufacture smooth surface parts made from polymers. Cordes, (2002), describes this process. This process starts with a master form of the required product which is immersed into a liquid preparation of the material and then allowed to dry. With repeated

immersions an amount of the liquid adheres to the product until the coating over the master form has reached the thickness to make the required product.

Textured surfaces and pre washing of the surfaces with blood have enabled successful transplants to take place.

A textured surface allows a layer of clotted blood to form so that the rest of the blood flows over the clotted layer and does not contact the material underneath. The initial blood flow is allowed to infiltrate into the pores of the textured surface and clot 'inside' the surface. This layer of clotted blood allows the rest of the blood to flow over the surface without any surface reactions. Figure 3.1 shows the result of one technique. As long as the clotted layer remains intact, the surface is safe to use.

Gagnon et al., (2000), indicate that with these techniques the HeartMate blood pump has shown a low incidence of thromboembolic complications, (3 – 5.2%), while the only patient anticoagulation treatment has been with aspirin and dipyridamole.



Gagnon et al, (2000). *Perfusion* 2000; 15: pp345-354.

Figure 3.1. Sintered titanium inner surface on HeartMate pump.
This surface promotes tissue ingrowth and prevents blood clots forming.

Powder metallurgy techniques are used to generate textured surfaces on metal parts by sintering 'microspheres' on to the internal blood contacting surfaces, similar to the HeartMate pump, and a textured surface technique can be used for silicone rubber parts. The sintering process is described by DeGarmo, Black et al., (1990).

Biocompatibility is not necessarily a property of the material itself. The Biomedical Engineering Handbook, (1995), indicates that the form and finish of the material is important and the geometry of the device is also important for biocompatibility in addition to the material used.

Some materials used for biomedical applications are porous. The size of the pore will determine the reaction between the device and the host tissue and are shown in Table 3.2.

<u>Pore size</u>	<u>Allows transmission of</u>
Less than 2 nm	Small molecules - gases
2 to 50 nm	Small proteins
Greater than 50 nm	Large proteins
Greater than 10^4 nm	Cells

The Biomedical Engineering Handbook, (1995).

Table 3.2. Relationship of pore size to materials transmitted.

Materials with the 'wrong' size pores can cause problems; an example is tissue ingrowth into artificial heart valves. Materials with the 'right' size pores are used where tissue ingrowth is encouraged; bone ingrowth into orthopaedic implants is an example. This topic is detailed in The Biomedical Engineering Handbook, (1995).

Many materials do not have a consistent pore size and if the material degrades, the pore size can change. This may affect the biocompatibility and the mechanical strength of the material. Brown University, (2002), indicates that the evaluation of biocompatibility involves the study of the damage and duration of effects on the host of any implant.

Biomaterials can be classified into basic types; these are metals, ceramics, polymers, composites and natural. Care is necessary to prevent the transmission of infections from the donor to the recipient when using natural

or biological materials. Much work has been done assessing these types for suitability. Table 3.3 shows some of the materials and applications

<u>Materials</u>	<u>Advantages</u>	<u>Disadvantages</u>	<u>Examples</u>
<u>Metals</u> Ti and its alloys, Co - Cr alloys, Stainless steels, Au, Ag, Pt etc.	Strong, tough, ductile.	May corrode, dense, difficult to make. Cannot sew through.	Joint replacements, bone plates and screws, dental root implants, pacer and suture wires.
<u>Ceramics</u> Aluminium oxide, Calcium phosphates including hydroxyapatite, carbon.	Very biocompatible, inert, strong in compression.	Brittle, not resilient. Cannot sew through.	Dental joint replacements, coating of dental and orthopaedic implants.
<u>Polymers</u> Nylon, silicone rubber, polyester, polytetrafluoroethylene etc	Resilient, easy to fabricate	Not strong, deforms with time, may degrade.	Sutures, blood vessels, hip socket bearings, ear, nose other soft tissues.
<u>Composites</u> Carbon - carbon, wire or fibre reinforced bone cement.	Strong, tailor made	Difficult to make	Joint implants, heart valves.
<u>Natural</u> Collagen taken from animals or other donors	Can be formed into almost any shape	May transmit infections, viruses to the patient	Heart valves, cosmetic reconstructions.

'The Biomedical Engineering Handbook', (1995).

Table 3.3 Materials for use in the body.

3.1.2 Mechanical characteristics.

The mechanical suitability of a material is governed by three main factors, the elastic modulus, the yield stress and the ultimate stress and these are described as follows.

a) The elastic modulus is the proportional deformation of the material within the limit of recoverable deformation as a result of stress applied to the material. Within the elastic limit the material can be loaded such that shape deformation occurs but when the material is unloaded, the material returns to the original shape.

b) The yield stress is the upper stress limit for a material without permanently deforming the material. Exceeding this stress will result in the material undergoing permanent deformation from its original shape and subsequently it may not possess the required strength to work satisfactorily. Depending on the characteristics of the material, failure may occur soon afterwards.

c) The ultimate stress defines the maximum strength of the material. At the ultimate stress the material has not failed but is permanently deformed. Any further loading of the material beyond this point produces a decrease in stress with increasing strain.

d) The fracture point is where the material fails completely and breaks.

Metals have a clearly defined elastic characteristic followed by a deformable region before the material fractures.

Ceramics have a very small 'elastic' zone. This is followed by the material cracking and then almost instant failure. It is difficult to determine the initial point where the material starts to crack.

Polymers have a very short elastic range followed by a very long, deformable, plastic range before failure. The material will not reform back to the original shape once the plastic range has been reached.

Examples of these characteristics are given in Okunade S, (2000) and DeGarmo and Black, et al, (1990).

Composites can be made with a combination of characteristics depending on the application.

3.1.3 Fatigue.

An important characteristic for all materials is fatigue failure. If a material is repeatedly stressed, eventually it will fail. The time to failure, or the number of times it is stressed before failure will depend on the material and the stress level. The stress level does not have to be close to the ultimate level for failure to occur.

3.1.4 Suitability of a biomaterial.

Materials that are super-hydrophobic, or highly water repellent, are being investigated to produce better water repellent coatings for clothing or windscreens. Sample, I. (2002), reports that developments have produced a material that does not contain any oxygen atoms, water is not attracted to it and the surface is covered in very fine ridges effectively reducing the area available for the water droplets to land on. BBC News (March 2003), report the development of a material treatment where water does not adhere to the surface. If this technology could be further developed, it may be possible to produce materials with a surface that would not attract or adhere to blood, as water is a major component of blood.

A better approach may be to produce materials that are actively 'antithrombogenic'. These will either encourage surface blood clotting to occur or to have active anticoagulant characteristics.

3.1.5 Physical and chemical problems.

Surface corrosion of a metal joint can increase the surface roughness and lead to increased wear. Particles wearing off the surface of the implant may cause it to become weak and fail.

If a protective surface is worn off or two dissimilar metals are in close proximity to each other then corrosion problems can occur. Corrosion is explained elsewhere in his thesis.

Surface fatigue fracture can lead to 'spalling' with relatively large particles of the material breaking off.

These are problems that must be carefully considered for every application of an implant.

3.1.6 Wear.

Wear is a problem with implants. For example, an artificial hip joint does not self-repair and with time, it may fail. But as the joint wears what happens to all the particles that wear away from the joint surfaces? The reaction by the body to these particles can cause problems and inflammation may occur local to the implant or at a site distant from the implant.

An implant does not have the capability to self repair during its working life and materials have to be chosen that are biologically and mechanically compatible to ensure the minimum wear possible between contacting surfaces. Examples of wear rates of different materials are shown in Table 3.4 and indicate about 2 magnitudes between the minimum and maximum rates of wear depending on the combination of materials.

<u>Material combination.</u>	<u>Material worn.</u>	<u>Rate (mm/yr)</u>
Metal / metal	Metal.	0.0003.
Metal / polymer	Polymer.	0.01 – 0.19.
Metal / composite.	Composite.	0.03 – 0.09 (estimated).
Ceramic / polymer.	Polymer.	0.008.
Ceramic / ceramic.	Ceramic.	0.003.

Medical Devices – materials for human contact. A design guide. (1996).

Table 3.4. Approximate wear rates of different material combinations.

Implants cannot be ‘serviced ‘ and combinations of materials that are ‘self lubricating’ are necessary. The metal / polymer combination for hip replacements have proved to be successful while blood lubricated bearings for the Jarvik 2000 axial flow blood pump appear to work satisfactorily.

3.2 Materials, metals.

Metals for implants are developed from structural and biological aspects. Early implants used to repair broken bones failed due to poor mechanical design and were made from materials that corroded.

There are at least six ways to prevent corrosion difficulties, some of which are not applicable to medical conditions. These are discussed below.

At least four distinct types of corrosion exist, these are described by Okunade, S. K., (2000).

A material suited to the conditions is important with medical implants. Although metals have sufficient strength and can be biocompatible, they have disadvantages that they are relatively heavy and a surgeon cannot attach them using the normal sewing techniques.

Metals can be heat treated to promote different metallurgical phases to give an improved corrosion resistance. Care is necessary not to promote worse conditions, such as grain growth, as larger interstitial spaces between the grains can be a source of a corrosion cell.

Heat treatment methods can be used but care is required in terms of quality control.

If a 'base' material is coated with a 'noble' material then the base is completely isolated from any electrolytic action. If the 'noble' coating is damaged, so that the base material is exposed, corrosion will start as the 'noble' material now acts as the cathode in a corrosion cell. (A 'noble' material is a material that is less electrically active than 'base' material in contact with it.).

With sacrificial methods, the anode corrodes in preference to the cathode and to ensure continuous protection, the anode is replaced at regular intervals.

While industrially this is a useful method, it would not be useful in medical applications for the following reasons

- a) Where would the sacrificed material go and will there be any effects on the patient from this material?
- b) How often would the anodes need changing to keep the system operating satisfactorily?
- c) How would the anode be changed if internally implanted?

Applying a voltage to base material will make it into a cathode. This may be possible in medical applications but difficult to maintain. The effects on the patient are unknown.

An example of an inhibitor is the antifreeze in a car engine. Chemical inhibitors in the antifreeze slow down the rusting process inside the engine. This process is not applicable to medical applications. Ruoff, A, (1973) describes this type of process.

Many of the above methods cannot be used in medical practice and the only defence against corrosion is to select the proper alloys and surface treatments for implants. For most alloys the corrosion rate is mainly dependent on the protective properties of the thin passive films that exist on the surfaces of these alloys. The quality of protection given by these films is related to their ability to resist chemical breakdown and their ability to reform rapidly once broken down, i.e., to repassivate. A report by Womato, I.H, (1999), covers this topic.

Ordinary steels are not suitable for medical implants. The development of stainless steels, with nickel and chromium, and cobalt chromium alloys has allowed a greater chance of success for this implant technique. The development of stainless steel type 316L, with a superior corrosion resistance to steel, is a good example of this. Appendix 5 lists the composition of 316L and the mechanical properties of surgical stainless steels. These materials are described in Parks et al., (1992), and De Garmo and Black, et al., (1990).

Hanson B., (1995) describes the uses and applications of titanium for metallic implants. Titanium has a high strength to weight ratio, good stiffness, a high heat transfer rate and is corrosion resistant. The properties of titanium are between those of aluminium and steel, with a yield strength of about 450 MPa, but this can be raised to about 1300 MPa by alloying and heat treatment. The cost of titanium is high in relation to steel due to fabrication difficulties and the high energy content required for manufacture.

Even with biocompatible materials, corrosion can still be a problem. If two materials of slightly different compositions, or even have areas of slightly different compositions, are in intimate contact with an electrolyte between them, it is possible that a small electrical potential will be set up. This can set up a small corrosion cell where one metal, the anode, will corrode away in preference to the other, the cathode.

This process can happen inside the human body with the body fluids acting as the electrolyte. It is important to use appropriate materials and avoid

implanting different materials in the same region even though they are both 'approved' materials. The manufacturer of medical implants has the responsibility to supply matched parts to ensure compatibility as far as possible and this responsibility can extend to the required grain structure of the material as, if the grain boundaries are large enough they can be considered to be small cracks that electrolyte can penetrate and initiate a corrosion cell. De Garmo and Black, et al., (1990) cover this in detail.

The failure of an implant is not always due to the materials used. As an example, if an artificial hip joint is poorly fitted or is overstressed by the user, the joint can fail through no fault of the implant itself.

When a new hip has been fitted, in order to give the implant a maximum possible life, patients are advised to follow some simple rules in order not to overstress the implant.

3.3 Materials, Plastics or Polymeric Materials.

Polymeric or plastic-like materials have some advantages in terms of the manufacturing properties as they can be moulded, cast, extruded and used as thin films or coatings. They can give a range of useful properties such as low density and good corrosion resistance, low tooling costs, and design versatility. They can also be manufactured to provide a controlled porosity. This can enable a device to be 'absorbed' into the body by means of tissue ingrowth or even act as a filter depending on the pore size. This is described in the Biomedical Engineering Handbook, (1995).

Correctly used these properties can lead to a product that is 'fit' for its purpose.

3.3.1 Structure.

The basic molecular structure of plastic is based on hydrogen and carbon in the relationship $C_n H_{2n+2}$. The hydrocarbons are linked to form very large molecules known as 'mers'. When many 'mers' are joined together a 'polymer' is created.

The same kind and number of atoms can combine in different structural arrangements and form different compounds with different properties. These

arrangements are known as 'isomers' and are similar to allotropism, (the property of existing in more than one form), or polymorphism, (any one of several forms in which the same thing may occur), in crystalline materials. This is described in De Garmo and Black, et al., (1990).

3.3.2 Biocompatibility problems.

Many types of plastics are used in the medical device industries for many applications but each has to be carefully selected. Some plastics can show poor biocompatibility, for example nylon absorbs water and can become soft and change shape.

Depending on the plastic, 'leaching' can be a problem where unwanted and potentially dangerous components of the plastic material itself appear in the blood or tissue of the patient.

This can have two effects,

a) The patient can experience discomfort as the unwanted material is absorbed.

b) The implant itself can degrade and lose strength or shape as described by Chandy and Sharma, (1991), and Costa and Maquis, (1998).

The reverse of leaching, that is absorption, can be a problem with some implants. Plastics are particularly prone to this effect. This happens when a component(s) of the body fluids pass into the material of the implant. The effects on the implant can be seen as swelling, distortion and it can lower the elastic modulus of the material and may lead to failure under physical loading.

Unwanted components of the material leaching out into the body cannot be accepted. Conversely the leaching characteristic can be used to advantage when planned leaching can be used as a drug delivery system to enable optimal doses of a drug to be delivered over an extended time interval as explained by Yamac T., (1996)

Applications for plastics include containers and piping as shown in Appendix 6.

3.4 Rubbers.

Silicone rubbers are used for the diaphragms of blood pumps and other products and have the advantages of low moisture absorption and are chemically inert but have to be carefully selected in terms of life and to avoid cracking. It is advantageous to build a diaphragm from multiple thin layers of material. This gives flexibility of the total diaphragm and added reliability as if one diaphragm layer fails the others will continue to operate.

3.5 Materials, Ceramics.

Ceramics could be used for the casings of artificial hearts, as they are hard, durable and show good biocompatibility. Unfortunately the broad application of ceramics is limited because they are brittle and are prone to sudden failure. They are therefore not used for parts of blood pumps that are heavily loaded as described by Doremus, (1992).

However with suitable designs this material has been used as part of the bearings of rotating parts of a blood pump, as the intrinsic hardness of ceramics gives good wear resistance and a long life.

There are additional problems with fixings using ceramic parts as obviously surgeons cannot sew through such a material and it tends to be heavy.

3.6 Materials, natural.

Artificial heart valves have been made using materials from animals where collagen from pigs has been used to make the flap part of the valve. However care must be taken to ensure the collagen is biologically 'dead' when used as it is possible to transfer pig cells to the human patient.

If cell transfer happens, it is possible to transfer viruses and other infections from one species to another. The effects of such a transfer are not known and may not be controllable once released. This topic was reported in Government press release (1998) and BBC News (July 1998) and Carlton TV, (2000)

Due to the possibility of rejection any natural or biological implants have to be very carefully selected and matched to the recipient as even transplants from members of the same family can be rejected. Various types of tissue transplants are described below.

Types of biomaterials.

Tissue transplants

A tissue transplant is where tissue is taken from a donor and transferred to a recipient. The donor and recipient may, or may not be, biologically related or even be the same species. The following transplant definitions are taken from the Online Medical Dictionary, http://www.online_medical_dictionary.org.

An autograft. Tissue is transplanted from one location to another within the same patient. A skin graft or the harvesting of arteries for coronary artery bypass are examples. There will be no rejection problems.

An isograft. Tissue is transplanted between genetically identical patients, (identical twins). There is a low possibility of rejection problems.

A homograft, or allograft. Tissue transplanted between members of the same species, for example a kidney transplant between two unrelated humans. There will be rejection problems and treatment will be necessary as the two patients will not be genetically identical.

A heterograft, or xenograft. A transplant between members of different species, for example a heart taken from a pig, and transplanted into a human. There will be rejection problems. This procedure is not used now as it was found possible that PERV, Porcine Endogenous Retro Virus, could be carried with the transplanted organ. The effect of this transfer across the species barrier was unknown and could not be estimated. This tissue transplant procedure is used for heart valves but the material of the valve has to be biologically 'dead' before transplantation into the patient and there is still the risk of rejection without treatment.

Non tissue materials.

These are man made materials used for repair or replacement purposes and have been covered elsewhere in this chapter.

3.7 Summary

This chapter has outlined the advantages and disadvantages of the materials used in blood contacting devices and has shown that care must be taken to avoid adverse reactions.

Some reactions can be used to advantage in tissue and bone repair and methods have been shown that can reduce adverse reactions. Manufacturing techniques are just as important as the choice of materials to promote biocompatibility. Surface texturing treatments can promote blood clotting to provide a blood compatible surface. Texturing can be applied to both metal and polymer surfaces.

There is not one material that is suitable for all applications. Each material must be chosen with the characteristics that are nearest to the 'ideal' for the application. For stressed parts, metals can be chosen as they have a predictable characteristics under this condition. Ceramics must not be used for stressed parts as instantaneous failure can result if over stressed or damaged. A combination of metals and polymers seems to give the best combination for bearing materials at present although ceramics, replacing the polymer, can be used on lightly loaded parts.

Materials that have the potential to cause cancer, or other problems, by material components leaching out into the body must not be used. Drug delivery systems using controlled leaching can be used but care must be taken to ensure the correct dose over the correct period of time.

Natural or donated tissue transplants would be ideal as they have the correct size and shape and perform the correct function. There are great difficulties with this process regarding rejection and the possibility of transferring infections and viruses from person to person or from animal to person. Some drugs exist to combat this effect but they are not completely effective.

The next chapter covers the design of the 'new' pump. Where possible existing theory has been used but if reliable theory has not existed a 'best judgement' approach has been used.

Problems with conventional pumps have been outlined and solutions offered. The design of the chosen pump type and the construction of the test rig are described together with instrumentation and calibration.

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Chapter 4, Design of the prototype pump.

4.1 Review of previous work.

From the discussions in Chapter 1, blood is easily damaged and it is important to select the 'best' pump for the application. A 'best' pump is considered to be a pump that delivers blood at the required quantity and pressure with little or no blood damage however it is caused. Some of the features necessary for a satisfactory blood pump are tabulated below.

<u>Advantages</u>	<u>Disadvantages</u>
Low blood shear stress	Large internal surface area.
Short blood dwell time	Incorrect materials and surface finish
Low anticoagulation treatment.	
No valves	Power transmission through the skin.
Acceptable blood temperature rise through pump	Unsatisfactory blood flow conditions include poor washing of surfaces, blood stagnation, blood cavitation, thrombus growth.
Simple control system	Installation problems with large pump

Table 4.1, Some of the advantages and disadvantages of a blood pump.

Some problems with blood pumps are discussed as follows.

Low pressure conditions can be set up in the blood by the blade of a rotary pump. If the pressure is low enough, cavitation can be set up, which can damage blood cells. Reul and Akdis, (2000), state that '*cavitation is most critical in axial flow pumps*' and that it is '*important to either monitor the inlet pressure or lower the rotation speed in case suction occurs*'. Any cavitation bubbles generated at the inlet are likely to pass through the pump and cause damage to the blood by the generation of emboli and possible damage to the pump and beyond. Reul and Akdis, (2000), also indicate that for rotary pumps, '*a maximum circumferential tip velocity of 10 m/s for any pump rotor*' is necessary to limit cavitation and blood shear rates. This implies that the smaller diameter axial flow pumps can run faster than radial flow pumps.

Whether this causes any additional problems in terms of bearing wear is not stated. At this limiting tip velocity, calculations show that a 5 mm diameter rotor can run at 38,200 rev/min while the speed for a 60 mm diameter rotor is 3,190 rev/min. The results of the calculations are shown on Figure 4.1.

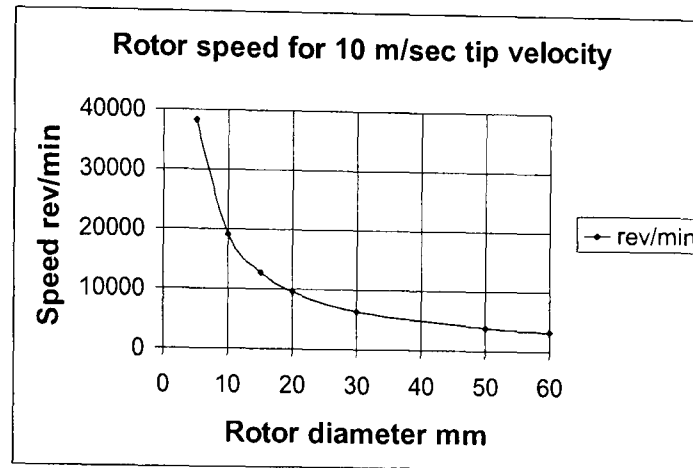


Figure 4.1. Rotor speeds for tip velocity of 10 m/sec.

Massey, (1995), comments *'that increasing the number of blades in a pump improves the flow conditions but decreases the pump efficiency due to blade friction with the fluid. Compromises have to be made in the number of blades to achieve the required results'*.

From this statement by Massey, the flow pattern in a pump with an infinite number of blades would be ideal but the efficiency would be low.

Applying these comments to a blood pump, a solution to the problems of pressure variations and shear around pump blades may be to use a rotary pump with no blades, just smooth disc(s). Morgan and Codispoti, et al., (1998), imply that cavitation would not be problem with this type of pump.

Hasinger et al., (1963), indicate that this type of pump *'while having a low efficiency exhibits a favourable cavitation characteristic'*. These are ideal flow characteristics for a blood pump if the low efficiency can be tolerated. Pumps of this type have been built before and are known as disc or shear force pumps

The above solution would be ideal for pumping blood although with large internal areas, the materials used to build the pump and surface finishes could be a problem in terms of blood compatibility.

Advantages of a disc pump.

Hasinger and Kehrt, (1963), put forward the following points regarding a disc or shear force pump

a) Shear forces are dependant on the difference in velocity between the fluid and the shear surface. There is no danger of fluid separation and a very stable pump operation should result.

b) The flow between the discs of this pump type is considered, generally, to be laminar. Only in larger rotors, with a disc diameter of approximately 300 mm and using water as a fluid is there a possibility of turbulent flow.

c) In comparison to the conventional pump the shear force pump appears favourable in terms of cavitation.

Disadvantages of a disc pump.

a) Due to the close spacing of the discs, the pumped fluid must be free from suspensions large enough to bridge the gap in between the discs.

b) At low flow conditions there is a tendency to trap gas bubbles in between the discs. If the ratio of the disc inner and outer diameters is kept small, say 2:1, this problem is reduced but conversely a disc with a small diameters ratio will reduce efficiency.

Following these examples, compromises will have to be made on each design and will be dependant on the application.

Tests were conducted by Rice, (1963), to determine the feasibility of multiple disc pumps and compressors using air and water as test fluids. The conclusions were as follows

a) *'Multiple disc pumps and compressors were feasible, they had advantages in special applications and were worthy of further investigation'.*

b) Disc turbo machinery efficiency was *'expected to remain constant as the machines were made smaller'*. This indicates that the disc pump may be superior to the conventional machine for very small volume pumping applications.

c) *'The disc pump may have an advantage when pumping 'exotic' fluids'.* These are fluids considered to have a density and viscosity greatly different from water.

d) *'The cost of disc machinery was considered to be low compared to ordinary pumps and that it could be manufactured in a modest machine shop'.*

e) *'The disc pump could be selected if, for a particular application, there is no advantage in performance between a conventional pump and a disc pump'.*

Calculated design data by Crawford and Rice, (1974), indicated that the *'multiple disc pump could be designed with high efficiency for fluid with any viscosity and density'.*

They also commented that *'it remained to be established by practice and investigation whether the pump has sufficient advantages compared to conventional machinery at the extremes of performance'.*

Part of the data presented by Crawford and Rice, (1974), was also suitable for gas blowers. One of the conclusions commented that *'if the pressure ratio was small enough, the speeds and stresses imposed on the discs are low enough that chemically inert materials, such as ceramic, could be used as disc materials'.* This would obviously be useful when pumping corrosive gases but would also be ideal for pumping blood as ceramics are biocompatible.

This paper assumed that the flow between the discs was laminar. It was stated that *'laminar flow exists for high values of the Reynolds number and low values of the volume flow rate parameter and for high values of volume flow rate parameter and low values of the Reynolds number. Turbulent flow exists for high values of both parameters'.* It is assumed from this statement that laminar flow results from low values of both the volume flow rate parameter and Reynolds number.

Laminar flow is preferred for a blood pump although small amounts of turbulent flow may be accommodated as there is evidence that a small amount of turbulent flow may exist in the aorta just after the blood leaves the heart.

From the above, it would seem likely that a disc pump would make a safe blood pump but with a relatively low efficiency compared to 'normal' rotary pumps.

4.1.1 Disc Pumps.

The history of a disc type pump goes back to 1850 when a water pump, invented by Sargent in the USA, was made from 29 circular discs mounted on a shaft with each disc spaced a few thousandths of an inch apart. This was the first known example of pump relying on the viscous drag / boundary layer principle. At this time it was considered to be unsuccessful and nothing was done with the idea. (Advanced Fluid Technologies, 2000)

A Serbian – American inventor, Nikola Tesla, took up this type of pump. Tesla, N, (1909), argued that a pump consisting of an assembly of circular discs rotating about a central spindle would pump fluid by virtue of *'two basic principles of physics, adhesion and viscosity'* and *'the disc pump, or shear force pump, uses the natural characteristics of adhesion and cohesion to move fluid'*. He argued that the device would accomplish this *by 'harnessing the internal forces opposing molecular separation' and 'the shock of the fluid against the asperities of the solid substance'*.

It is assumed that Tesla was indicating the fluid adhered, or was sticking, to the surface of the discs. The work of Poiseuille with flow through tubes indicates this characteristic where he described that the 'layer' of fluid closest to the wall of the tube remained stationary while the body of the fluid was moving relative to the tube. If the walls of the tube are 'replaced' by the surfaces of discs which then rotate, the fluid will tend to rotate with the discs.

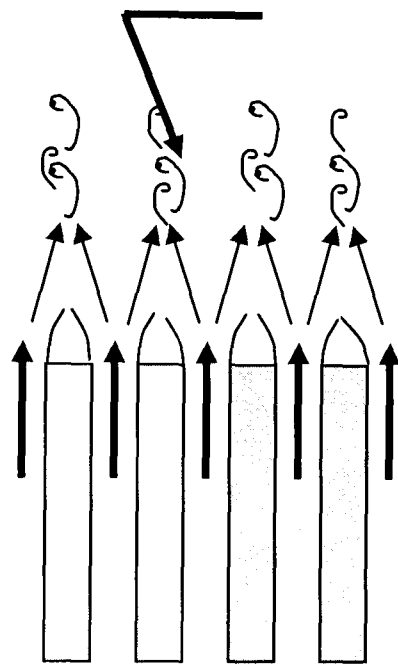
This implies that the fluid was 'sticking' to the surface and deformities of the discs, even though the discs were as smooth as possible and the 'deformities' may not be intentional.

Massey, (1995), describes the effect of fluid 'sticking' to a surface as the 'boundary layer' which is the layer of fluid closest to, say, the wing of an aircraft. An amount of fluid will be drawn along with the wing. This can create a 'wake' of disturbed fluid that can last for a considerable distance after the

wing has passed. The length and energy in the wake will depend on the shape and speed of the wing.

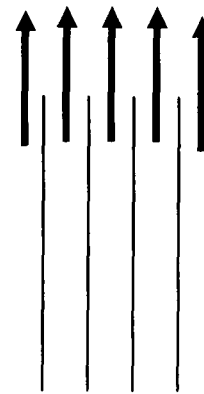
Applying this 'wake' concept to disc pumps, either the discs will have to be made of material that is 'thin' relative to the diameter or the outer edges of the discs will have to be tapered, (using an ogival shape), to reduce the wake formation and turbulent flow problems. Figure 4.2 illustrates the problem and possible solution.

Large turbulent flow region
in 'wake' after discs.



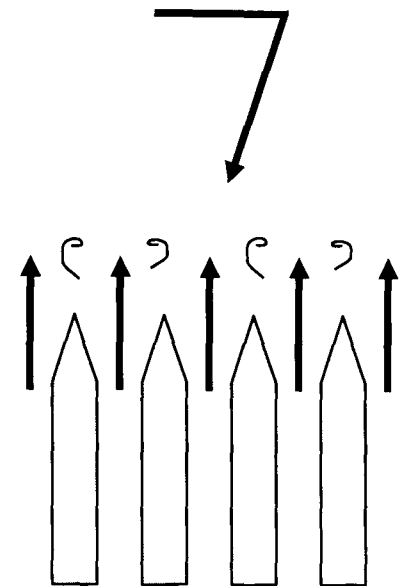
Thick discs, blunt edges.

Smooth flow,
little turbulence.



Thin discs

Smooth flow,
little or no turbulence.



Thick discs,
tapered edges.

Figure 4.2. Effect of blunt edges, thin discs and tapered edges
on fluid flow through the disc pack.

With cohesion the molecules of a fluid are attracted to each other at the molecular level, Van der Waals forces are an example of this.

Van der Waals forces and cohesion are part of the physical forces of attraction and repulsion existing between molecules and are responsible for the cohesion of molecular crystals and liquids. On a practical level, once part of the fluid has started to move, for example the fluid immediately adjacent to the discs, the rest of the fluid will tend to follow.

In 1909, Tesla applied for patents for a turbine and a pump based on this principle. Figure 4.3 shows part of the patent application for the pump.

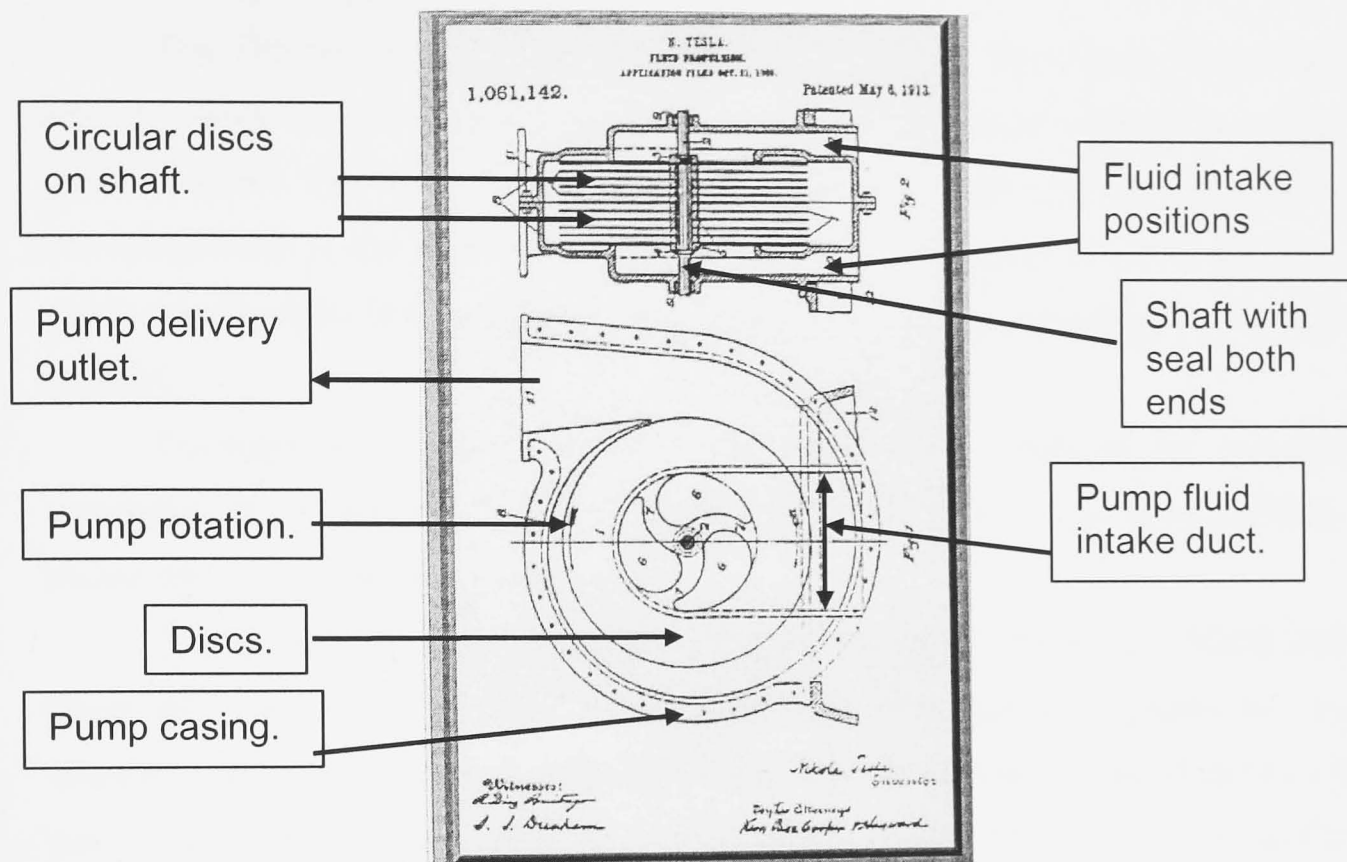


Figure 4.3. Tesla Turbine Pump.

Part of the drawings submitted for US Patent no. 1,061,142

by Nikola Tesla in Oct. 1909.

Source : - The Animated Software Company, Internet Glossary of Pumps. <http://www.animatedsoftware.com>

There appears to have been little or no commercial exploitation in this design of pump until the 1970's when a Californian inventor, Mr. M. Gurth, took it up. He found that when increasing the disc spacing to as much as 500 mm, the boundary layer / viscous drag principal still applied and the pump worked well especially with liquids of a high viscosity, (250 cP or 0.25Ns/m² and above). Gurth took out US patents in the late 1970's and founded the Discflo Corporation in 1982. Further developments of this pump have since been patented.

This type of pump is used to move viscous, abrasive fluids, fluids with high solids content, gas / air entrained fluids and delicate and shear sensitive products. The Discflo Corporation claim to pump living cells, plankton, bacteria, live fish, creams, liquid sugar, grapes, whole chickens. This type of

pump is claimed to be more reliable with lower life cycle costs than other pumps for this type of application. They appear to be the only company in the world making this type of pump. (Discflo Corporation)

The Discflo pump is the same basic design as the Tesla pump built with a series of flat discs, separated so that there is a clearance space between them. The first disc is mounted to a driving shaft and the other discs are connected to the first disc. The disc pack is contained within a housing, which has the inlet at the centre of rotation and the outlet tangential to the disc rotation.

Damage and wear with this type of pump is said to be minimal compared to a normal type of pump with blades on the disc surface. (Advanced Fluid Technologies, 2000).

Evidence to promote this type of pump comes from Rice, (1963) and Hasinger and Kehrt, (1963), who considered this type of pump as an *'excellent pump for exotic fluids due to the inherently stable flow regimes'*. It has been successfully used to pump liquid fuels for the NASA space programme.

Some work has been done by Miller et al, (1989), and Miller et al, (1990), as a pulsed output pump. The pump used polycarbonate discs mounted on a shaft. Each disc had an outer diameter of 76.2 mm, (3 inches) and an inner diameter of 38.1 mm, (1.5 inches). The discs were 1.6 mm, (0.063 inches), thick and they were spaced 0.406 mm, (0.016 inches), apart. The fluid entered the pump through the rotor pack at the opposite end to the driving shaft and as the fluid entered the spaces between the discs, it was sent into a spiral motion with the radial and frictional forces generated by disc contact. As the fluid spun off the surface of the discs, it was collected by the spiral volute chamber, acting as a housing and diffuser. The fluid then left the pump tangentially through the outlet port built into the housing.

There was no evidence of any output performance results.

Yamane et al, (1996), and Jarvik, (1995), have indicated that a centrifugal type rotary pump was more suitable than an axial flow pump for artificial heart applications since a rotary type pump operates at larger clearances between the rotor and the housing than an axial flow pump giving less chance of shear and thrombus problems.

The suitability of a disc type rotary pump can be further argued by considering the flow of blood through a pipe when the red cells in the blood move towards the centre of the pipe leaving a layer of plasma at the pipe walls. The mode of operation of this pump is very similar if it is considered that this time it is the 'pipe', (the discs), that are moving and the blood is being drawn along with them. This implies that this pump type 'pulls' the product smoothly and without pulses through the system compared to other pump types that 'push'. If the red cells are kept away from the surface of the discs, this implies a reduced chance of mechanical damage to the cells through collisions with the walls and therefore a reduced chance of haemolysis and the formation of blood clots. Figure 4.4 outlines the expected velocity profile.

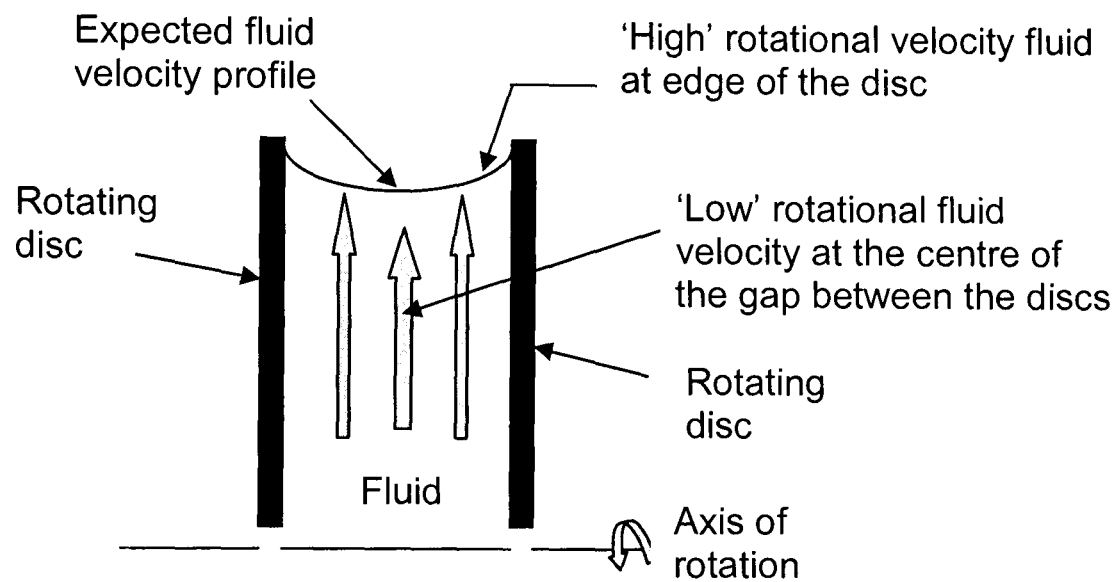


Figure 4.4 Expected fluid velocity profile between the discs of a Tesla type pump.

An approximation of the fluid velocity profile in between the discs can be given if the fluid is considered as a series of layers. Using the previous explanations involving fluid flow in tubes, as the fluid enters the pump at the centre of the discs, the 'peripheral' velocity at this point will be low depending on the radius of the hole.

As the fluid is accelerated circumferentially by the rotation of the disc, the fluid will move outwards radially across the face of the disc. The fluid layer closest to the discs will be moving the fastest of all the layers as it has been subjected to the greatest shear force. This layer will continue accelerating

circumferentially until it reaches the same peripheral velocity as the disc or leaves the disc surface at its outer radius. The adjacent layers will move at a slower speed depending on the viscosity of the fluid and the distance away from the disc surface. It is assumed that the velocity profile between two adjacent discs will tend to be a parabolic shape, with the slowest flow rate in the middle of the two discs, similar to the pipe flow profile.

As there are no blades moving the fluid, there is no chance of large pressure changes and therefore little chance of cavitation.

There is no information in terms of actual, measured, internal flow patterns and pressures for this type of pump. This is probably due to the practical difficulties of fitting probes inside the rotating mass of fluid without affecting the characteristics of the fluid flow.

Papers by Leaman, (1950), Armstrong, (1952), Young, (1957), Smith, (1960) and Gordon, (1962), suggest increased interest in this type of pump at those times. All these papers were referenced from Crawford and Rice, (1974)

4.2 Design choices.

Attempting to build one of these pumps presented problems. For pumps with blades there was information on blade angles, velocity diagrams etc. in many textbooks, but there was no information about disc pumps. Enquiries from various manufacturers did not provide useful answers. The company that manufactures disc pumps, the Discflo Corporation, were only willing to supply sales and performance information.

Figure 4.3 indicates the discs were made from thin section material, which is material that is thin relative to the disc diameter. Measurements taken from the drawing indicate a disc diameter-to-thickness ratio of around 55 : 1. If this ratio is strictly followed, the manufacture of a 'small' pump will be difficult as it may not be possible to obtain biocompatible materials with enough physical strength that are thin enough and flat enough to maintain position and prevent collapse when the pump is stationary. A 'small' pump using thin discs will need to run at high speeds for the required output.

Stresses on the discs will have to be carefully assessed to ensure that the discs will not 'grow' due to creep and the disc will not break up during use.

It was therefore considered that the pump produced would not be ideal due to the lack of reliable design information.

In order to make the pump simple, an overhanging rotor design was used. With type of design, the rotor is located outside the bearings and is typical of many commercial designs. Figure 4.5 shows the general arrangement. This requires a shaft seal to prevent fluid leaking out of the pump. For a blood pump this is not a good feature as blood becoming trapped between the shaft and the seal will be damaged and the friction of the seal rubbing on the shaft will cause heat to be generated that has to be dispersed in the blood. From previous work, the maximum temperature that can be withstood by the skin is around 40 / 41° C. and blood is expected to have a similar value before being damaged. As blood would not be used for these tests this feature was considered satisfactory.

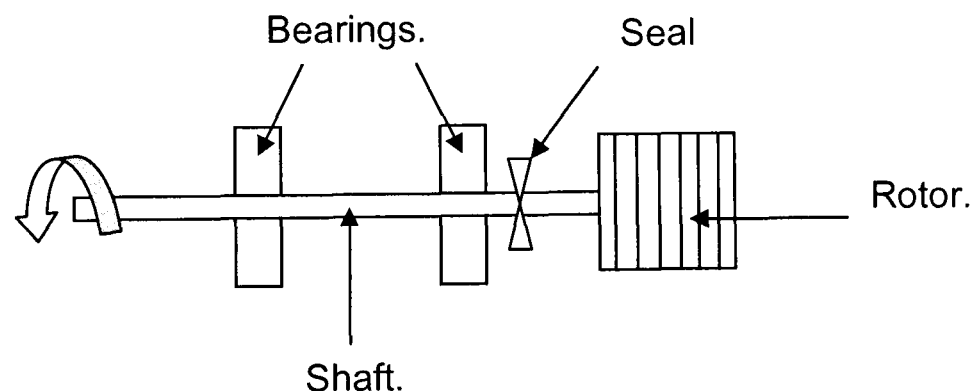


Figure 4.5. Outline of pump design.

The pump body would be built in three sections and clamped together with through bolts. Section 1 would hold the shaft, bearings and seals, section 2 would be the volute containing the rotor and section 3 would be a fluid inlet section. Although there might be problems sealing the pump sections, this design would make it easy to change the rotor.

Figure 4.6 outlines the basic construction of the pump.

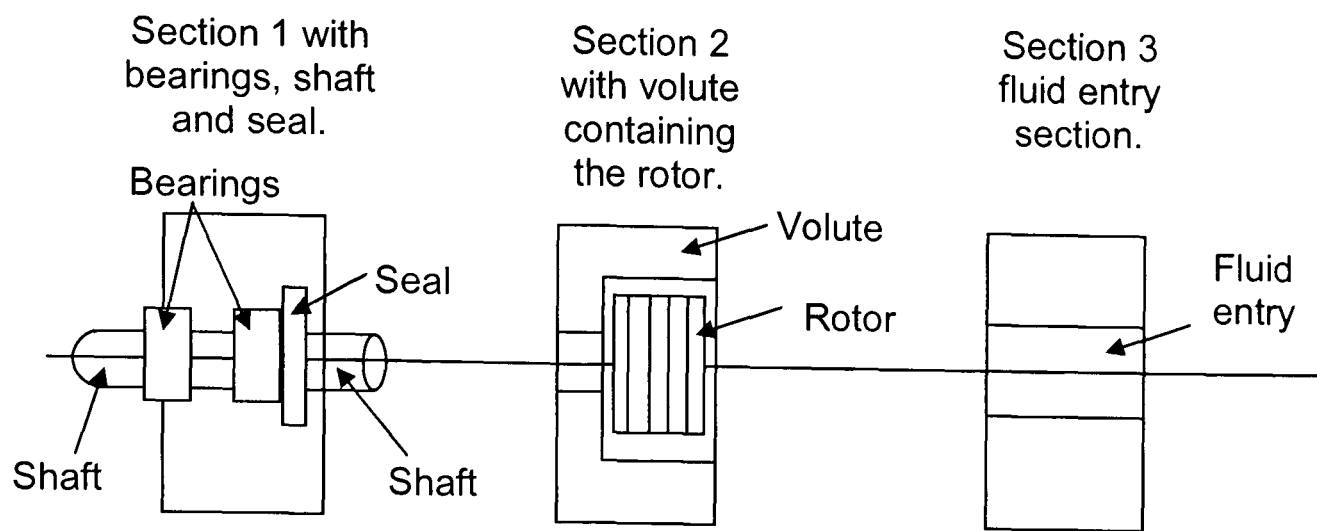


Figure 4.6 Layout of the sections of the pump.

4.2.1 Material.

A transparent material was required for the discs and pump so that internal flow patterns could be observed. Perspex was chosen as it is commercially available, it can be machined easily and can be polished to obtain a 'see through' surface.

The thickness of each disc was to be as small as possible in order to pack as many discs as possible into a rotor assembly. The thinnest material commercially available was found to be 3mm thick, anything thinner than this would be a 'special' order and the thickness and flatness were not guaranteed.

4.2.2 Discs and rotor assembly.

It was not known if there was a 'best' ratio of the outer diameter and the diameter of the hole through the disc. It was considered necessary that the hole in the centre of the discs was a similar diameter to that of the human aorta. From Grays Anatomy, (1995), the diameter of the aorta as it emerged from the heart is given as around 1.25 inches, (31.75 mm), and to be approximately circular in section. These figures are not exact as people vary slightly. The measurements from Figure 4.3, the Tesla turbine pump, gave a disc diameters ratio of around 1.8 : 1.

The hole through the discs was chosen as 30 mm diameter and the outer diameter as 60 mm, giving a disc diameters ratio of 2 : 1. This will keep

the pump to a reasonable overall physical size considering the possibility of development into a marketable product. The disc diameter ratio may not be the best for pump efficiency but the performance could be used for comparison purposes.

The two end discs were to be thicker in section than the rest of the discs. This would give a location for the through bolts that were going to be used to hold the disc pack together.

The rotor design was in three parts and the assembly of the discs was to be as follows.

The first disc, the mounting disc, was to be fixed to the shaft through a square hole in the disc fitting over a square section spigot on the mounting boss fitted to the shaft. The hole in the centre of the spigot was threaded and a bolt was used to clamp the mounting disc into place on the spigot. The driving discs were to be assembled on the mounting disc, using through bolts and spacers, until the required number of discs had been fitted for the assembly. The last disc of the rotor, the end disc, was to be assembled in the same way as the driving discs using the through bolts.

The total width of the disc pack, or rotor, was to be set so that a clearance of 1.5 mm was maintained between the end discs and the internal walls of the volute for all the rotor builds. The clearance was determined by knowing the thickness of each disc, multiplying by the number of discs, adding the total of the gaps between each disc and subtracting this from the width of the volute. Dividing this figure by 2 gave the calculated clearance.

The material for the discs was thicker than ideal and it was considered that the flow patterns near the edges of the discs would not be ideal. Tapering the edges of the discs was considered but the time involved was considered to be excessive in terms of the machining and polishing involved.

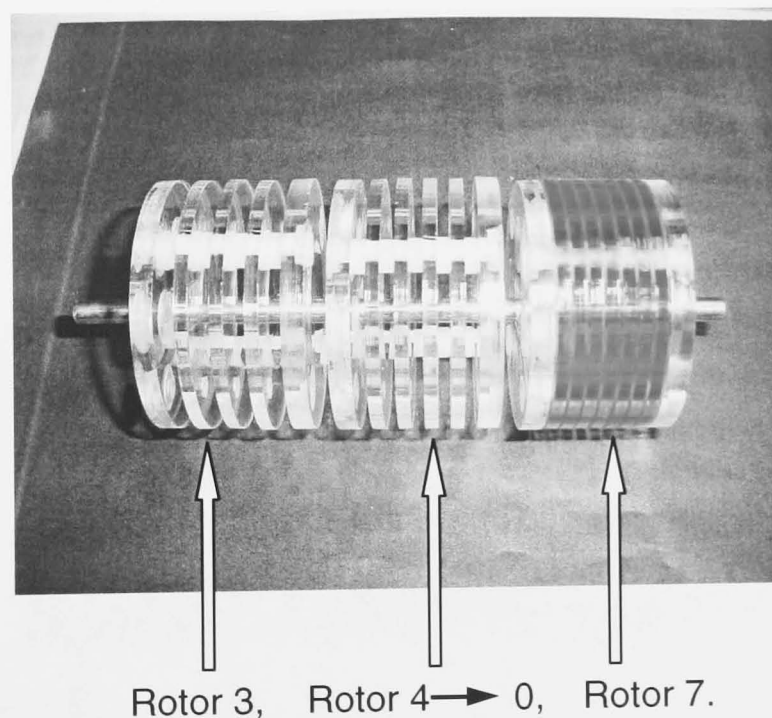
Photograph 4.1 shows the three rotors built for the tests. The rotor on the left was built with three pumping discs and was named rotor three. The middle rotor was originally built with four pumping discs but was later modified so that all the pumping discs were removed leaving just the end discs. This rotor was named as rotor zero. It is shown in Photograph 4.2. It was intended to determine if a pumping action was possible with this arrangement.

The rotor on the right in Photograph 4.1 had seven pumping discs and was named rotor seven.

The surface area, mass and moment of inertia were calculated for each rotor and are listed in Figure 4.7. The mass and moment of inertia for the electric motor armature were estimated assuming the armature to be made of steel, (density 7,850 Kg/m³), and to be 50 mm diameter and 100 mm long.

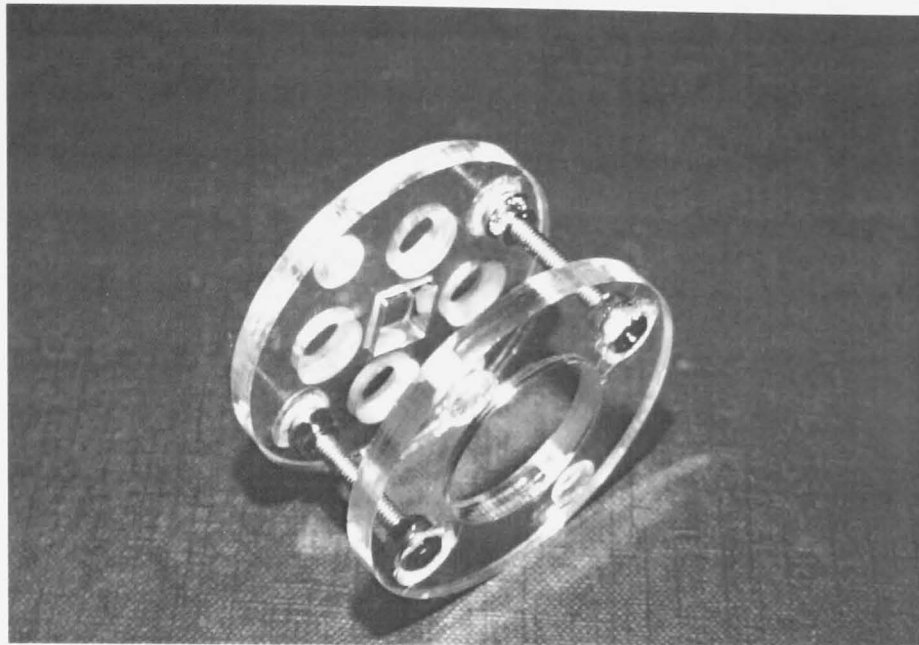
	<u>Surface area m²</u>	<u>Mass Kg</u>	<u>Moment of Inertia Kg.M²</u>
Rotor 0	0.0383	0.037	0.00001326
Rotor 3	0.089	0.059	0.00002520
Rotor 7	0.157	0.084	0.00004083
Motor armature	-----	1.541 (est.)	0.0004816 (est.)

Figure 4.7. Calculated attributes for the pump rotors and the motor armature.



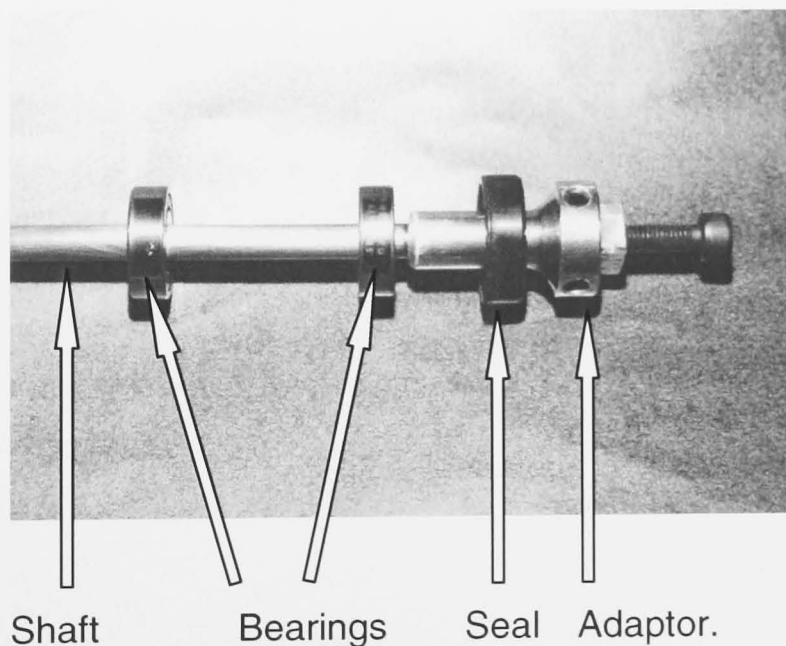
Photograph 4.1. The three rotors as originally built.

The gap between each disc is as follows, rotor 0, 24 mm, rotor 3, 3mm and rotor 7, 1 mm.

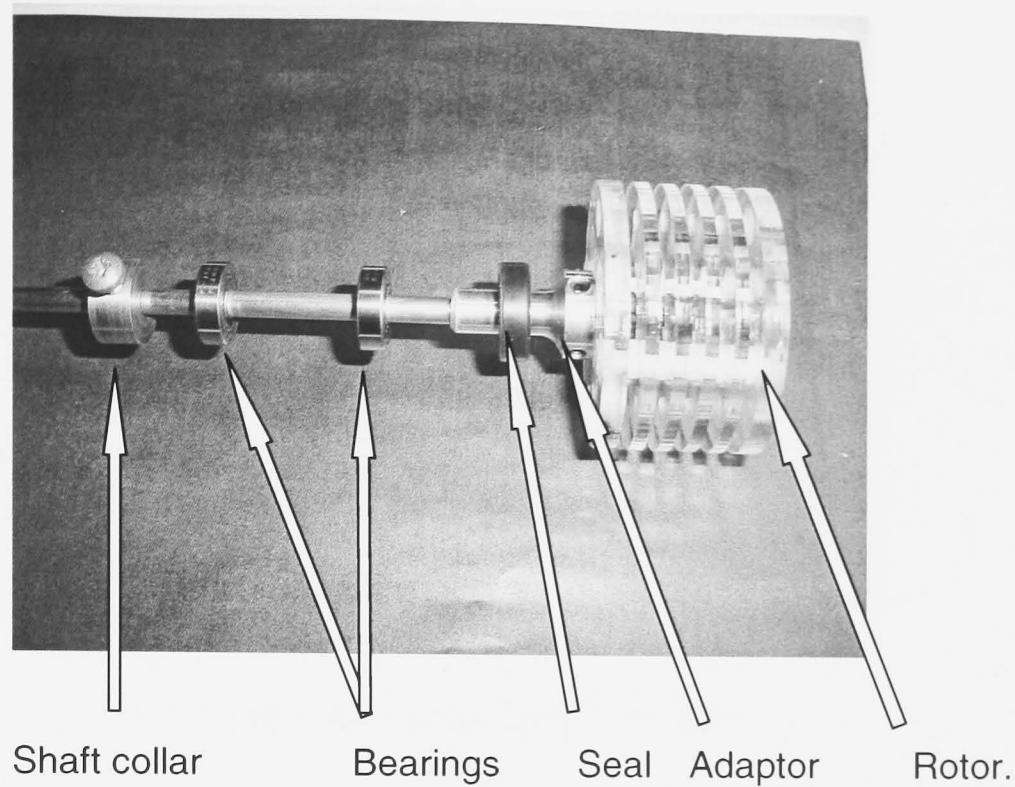


Photograph 4.2. Rotor 0 - two end discs only.

Photograph 4.3 shows the arrangement of shaft, bearings, seal and adaptor. This shows the adaptor fitted to the shaft and how the shaft was carried on two ball bearings. The seal was arranged to prevent fluid leakage out of the pump. The rotor fitted on to the adaptor as shown in Photograph 4.4.



Photograph 4.3. Shaft with two bearings and seal on adaptor.

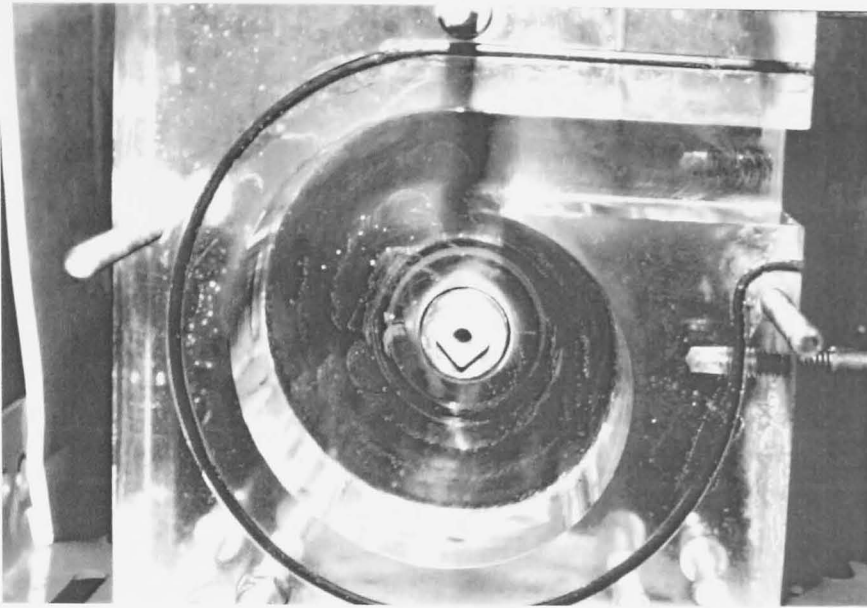


Photograph 4.4, Shaft with a rotor in position on adaptor.

The shaft collar was to be used to prevent shaft end movement when the pump was running.

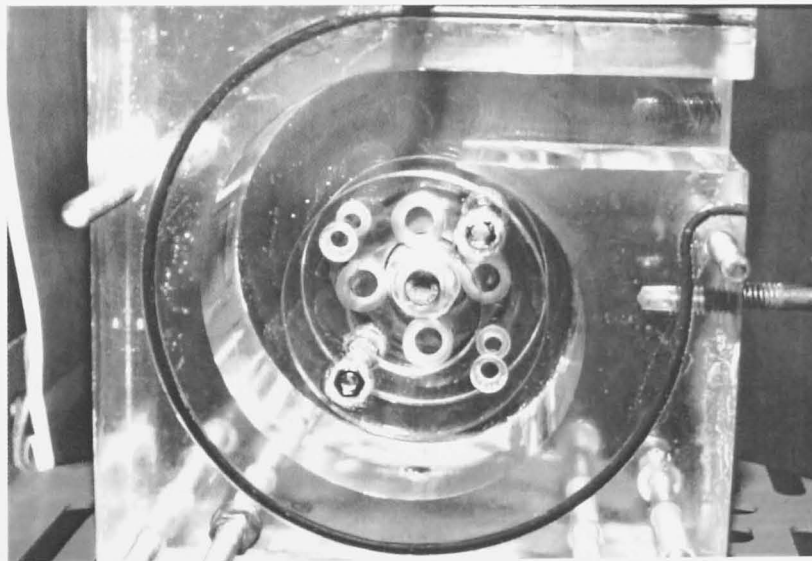
The end of the shaft was left plain and was connected to the motor by a flexible coupling to compensate for any 'out of alignment' with the motor and shaft when assembled into the test rig.

Photograph 4.5 shows the pump without the rotor fitted. The shaft and adaptor are in place and the square drive on the end of the adaptor can be seen in the centre of the volute.

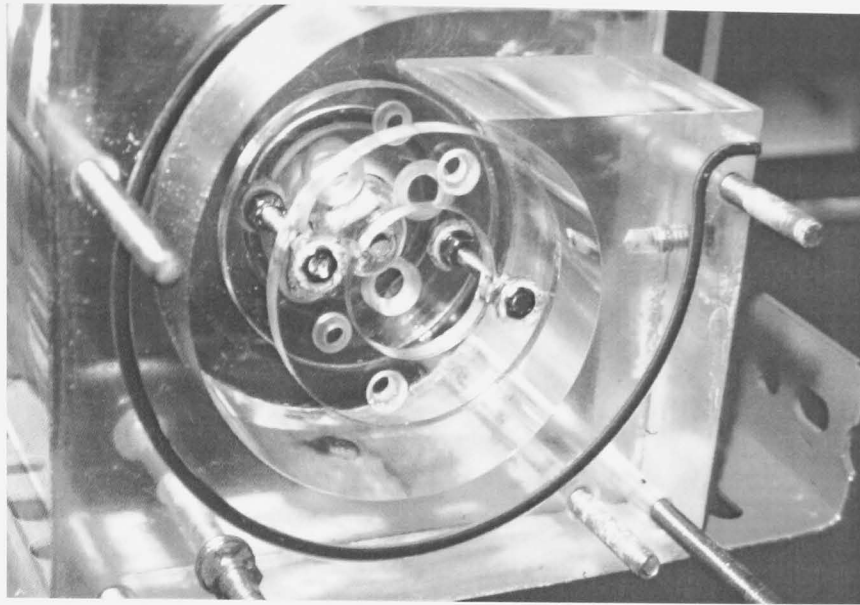


Photograph 4.5. Shaft fitted into pump.

Photographs 4.6 and 4.7 show the rotor fitted on to the adaptor and where the rotor runs inside the volute.



Photograph 4.6. Rotor zero fitted to adaptor and located inside the volute.



Photograph 4.7 Second view of rotor zero assembled to the pump.

4.2.3 The pump volute.

The internal shape of a pump volute is ideally a logarithmic spiral. From the equipment available at the University it was not possible to produce this shape. It implied many small segments joined end to end to form the logarithmic curve which would have taken considerable time and be open to mistakes. An alternative method used two half circles on different centres. This gives a shape close to a logarithmic spiral but using the standard routines built into the machine is easier to make and be less liable to mistakes.

The dimensions for the volute are related to the thickness of the material used and the outlet area of the volute. These were taken into consideration in the volute design.

In order to allow a clearance between the edges of the rotor discs and the cutwater point in the volute, the distance from the rotor centre-line to the cutwater point was set at plus 10% of the rotor outer radius, (33 mm), as detailed in Turton, (1993), Figure 4.8 shows this arrangement.

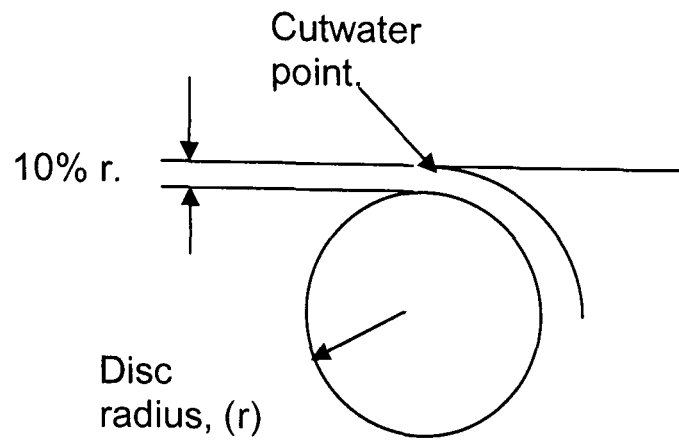


Figure 4.8 Detail of cutwater point.

The pump outlet was to have the same area as the aorta as it leaves the heart. Grays Anatomy, (1995), gives this as equivalent to 791.73 mm².

Material, 50 mm thick, was chosen for the pump volute in order to a) fit the rotor in position but b) to provide enough additional material to provide a structural restraint for the volute and prevent it 'springing' open after machining. Suitable material was purchased from a supplier. Given the required outlet area and material thickness, a volute design was made based on these figures.

The calculations for the volute dimensions are shown on Table 4.2.

The drawings for the pump parts are shown on Appendix 10.

Table 4.2, Volute calculations.

Diameter of human aorta at heart exit	31.75 mm	Area of human aorta	791.73 mm ²
Width of passage	40 mm	Cutwater radius	33 mm

<u>Degrees</u>	<u>Percentage</u>	<u>Volute area</u>	<u>Volute depth</u>	<u>Volute radius</u>	<u>Volute tube</u>	<u>Volute tube</u>	<u>Tube 'rad'</u>
<u>around volute</u>	<u>around volute</u>	<u>mm²</u>	<u>mm</u>	<u>depth</u>	<u>area. mm²</u>	<u>dia.</u>	<u>+ 33 mm</u>
0	0.00	0	0	33.00	0		
30	8.33	65.98	1.65	34.65			
60	16.67	131.96	3.30	36.30			
90	25.00	197.93	4.95	37.95	197.93	15.875	40.938
120	33.33	263.91	6.60	39.60			
150	41.67	329.89	8.25	41.25			
180	50.00	395.87	9.90	42.90	395.87	22.451	44.225
210	58.33	461.84	11.55	44.55			
240	66.67	527.82	13.20	46.20			
270	75.00	593.80	14.84	47.84	593.80	27.496	46.748
300	83.33	659.78	16.49	49.49			
330	91.67	725.75	18.14	51.14			
360	100	791.73	19.79	52.79	791.73	31.750	48.875

4.2.4 Summary.

The disc pump appears to have favourable characteristics when applied to pumping blood. These include low shear stresses and the ability to handle fluids with viscosities that are greater than water. The ability to safely handle fluids with inclusions is also favourable.

The pump is simple to build and if the disc stresses are low enough, ceramic materials can be used to fabricate the discs. This point is important as ceramics are biocompatible and do not react with blood.

If the pump is built with 'thin' section discs, there should be little or no problem with disturbed fluid flow patterns as the fluid leaves the rotor.

A disadvantage with this pump is the efficiency which may not be as high as a conventional pump. If the disc pump is designed for a particular application, the difference may be small compared to a conventional pump, and may be acceptable if an alternative pump is not available and the advantages outweigh the disadvantages.

Another disadvantage with this pump design is the large internal area available for blood contact with the multiple discs. Careful selection of a biocompatible material and design will reduce this effect.

The next chapter covers the building of the test rig and the testing of the pump.

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Chapter 5, Pump testing.

5.1 Test Rig, Introduction.

A rig was constructed to test the pump. The pipe work was commercially available 25 mm, (1 inch), bore plastic pipe with gate valves fitted at the pump inlet and outlet to control the flow and pressure of the pumped fluid.

Instrumentation was used to measure the pump speed, fluid flow, strain and differential pressure across the pump inlet and outlet, together with the power required to drive the pump. A brief description of the equipment and operation together with photographs of the rig and pump as installed are shown later.

Electronic strain gauge differential pressure transducers, Radio Spares part no 286-686, were used to measure the pressure generated by the pump. The transducers, maximum pressure difference of 34.5 kN/m^2 , (5 lbf/in^2), were calibrated using a mercury manometer to determine the relationship between pressure and strain.

Figure 5.1 shows the pressure calibration equipment used.

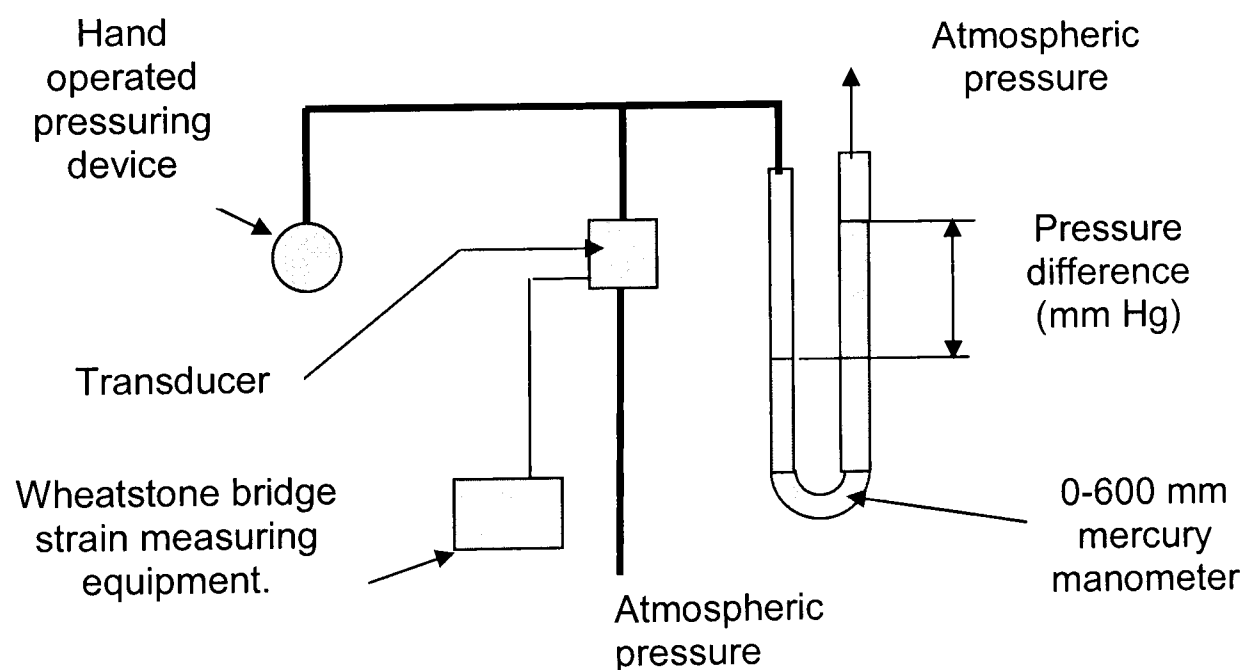


Fig. 5.1. Pressure calibration equipment.

The results are shown on Appendices 7.1 and 7.2. The characteristic between pressure and strain was a straight line relationship. The slope of the

line was calculated as 0.02883 and the formula below was used to calculate pressure from the measured strain.

$$\text{Pressure (mmHg)} = 0.02883 \times \mu\text{strain}$$

Pressure tappings at the inlet and outlet of the pump were connected to the pressure transducers. The pressure was calculated using the formula above.

Before every test, the output from each transducer was zeroed to ensure that the pressures measured were accurate relative to atmospheric pressure.

In order to drive the pump, an electric motor and associated control equipment were purchased from suppliers. Details are shown in Appendix 8.

Tests were conducted to determine the tacho generator output voltage relative to the tacho generator speed. By plotting this characteristic, an equation was obtained which was a close match to the plotted results as shown in Appendix 9. Using this equation the pump speed was calculated from the tachogenerator voltage.

It was considered fitting thermocouples to the pump inlet and outlet to measure temperature differences. A digital thermocouple was used to measure the temperatures. The inlet and outlet of the pump were considered to be suitable temperature measuring points and a small quantity of Vaseline petroleum jelly was placed at these positions. The temperature probe could then be inserted into the Vaseline during the test and it was assumed that the Vaseline was at the same temperature as the point to be measured.

A test was taken to determine temperature rise using water as the fluid. The pump was set to deliver 1 l/min at a differential pressure of 200 mm Hg. This resulted in an average pump speed of 4013 rev/min. The temperature was taken every 15 minutes for 5½ hours. Each reading was at an accuracy of +/- 0.1 °C

A temperature rise of 16 degrees C over this time period was noted. The room ambient temperature was about 20 – 21 degrees for the duration of this test.

A plot of the results is shown on Appendix 11

A Rotometer with a flow range of 0 – 10 litres/min was used to measure fluid flow. (Elliott Process Instrumentation, SM 18 S-1). This is a commercial device and operates on the principle of drag on a submerged body.

5.2 Pump rotors.

Three rotors had been built and are described in Chapter 4.2.2 and are shown in Photographs 4.1 and 4.2

The rotors were to be tested in the following order.

- a) The rotor with two end discs only
- b) The rotor with seven discs plus two end discs
- c) The rotor with three discs plus two end discs.

This order was assumed to be the most suitable in terms of performance and it would hopefully leave the pump built with the optimum rotor at the end of the tests.

Tests were conducted and the details are shown in Chapter 5.6.6 onwards.

After the tests with the first rotor, the water was drained out of the test rig and the rig was run through a full speed range from 'fast' to 'slow' and back to 'fast' again in order to determine any hysteresis. Results were plotted showing tachogenerator volts and motor power consumption, (watts) as shown on Table 5.1.

The characteristic equation for this relationship was determined and is shown below.

$$P = a_0 + a_1N + a_2N^2 \quad (1)$$

Where P = power (Watts)

N = speed (Rev / min)

$$a_0 = 2.53$$

$$a_1 = 7.866 \times 10^{-3}$$

$$a_2 = 5.411 \times 10^{-7}$$

From this it was possible to calculate the no load power consumption at any speed. Subtracting this power from the power recorded during the tests would give the actual power required to pump the water.

Tacho. output volts.	Speed, rev/min.	Average no load power, Watts
20	3928.3	36.46
18	3440.5	31.08
16	2974.9	26.09
14	2531.6	20.95
12	2110.5	16.24
10	1711.7	12.19
8	1335.1	8.77
6	980.9	6.05

Table 5.1, Results of speed and 'no load' power consumption test.

5.3 Physical similarity, introduction.

It is important to predict product performance in order to prevent an unsatisfactory product reaching the market place. It is not always possible to test a full size prototype and model tests are used to predict the performance of the prototype. The model results are then 'scaled up' to obtain the results for the prototype.

For the results to be valid the model must be '*physically similar*' to the prototype. This term covers several kinds of similarity and these are described in Massey, (1995). In all these descriptions it should be understood that two systems are physically similar '*when the ratio of the corresponding magnitudes of these quantities between the two systems is everywhere the same* '.

5.3.1 Non dimensional performance of pump.

Non-dimensional coefficients can be used to predict the performance of machine B from the results of machine A as long as they are both geometrically similar machines. In order to provide a comparison with other pumps from published papers, these have been applied to the results obtained from the Tesla pump tests.

Formulae for the non dimensional coefficients have been taken from Crawford and Rice, (1974), and are listed below together with the symbols used.

$$\text{Efficiency } \eta = \frac{Q \bar{p}_t}{P} \quad (2)$$

$$\text{Reynolds number } N_{RE} = \frac{\rho \Omega h^2}{\mu} \quad (3)$$

$$\text{Volume flow rate parameter } U = \frac{Q}{2\pi r_i^2 \Omega h} \quad (4)$$

$$\text{Dimensionless torque } T_1 = \frac{P}{N \rho h \Omega^3 \bar{r}_i^4} \quad (5)$$

$$\text{Dimensionless pressure change } p_t = \frac{\bar{p}}{\rho \Omega^2 r_i^2} \quad (6)$$

An explanation of the symbols is shown in Table 5.2. The figures for rotor 3G refer to the density and viscosity of the water / 45% glycerine mixture and are a straight line interpolation from Kaye and Laby, (1995) and are shown as *. The figures taken from Lönn, (1997) are shown as **.

The range of values for the non-dimensional coefficients quoted by Crawford and Rice, (1974), are as follows.

Efficiency η	0 to 99
Reynolds number N_{RE}	0 to 10
Volume flow rate parameter U_o	0 to 2
Dimensionless torque T_l	0 to 1250
Dimensionless pressure change p_t	3.5 to -437

<u>Symbol</u>	<u>Description.</u>	<u>Rotor 0</u>	<u>Rotor 7</u>	<u>Rotor 3</u>	<u>Rotor 3G</u>
Q	Flow through entire pump.	m ³ /sec	m ³ /sec	m ³ /sec	m ³ /sec
p_{ut}	Pressure change through pump	Pa or N/m ²	Pa or N/m ²	Pa or N/m ²	Pa or N/m ²
P	Power supplied to pump shaft.	Watts	Watts	Watts	Watts
ρ	Density of water (kg/m ³)	1000	1000	1000	
ρ	Density of test fluid				1100 * 1055 **
Ω	Rotational speed.	Red/sec	Red/sec	Red/sec	Red/sec
h	Spacing between discs (m)	0.024	0.0005	0.003	0.003
μ	Viscosity of water (kg/m.s @ 20 deg C)	0.001002	0.001002	0.001002	
μ	Viscosity of test fluid.				0.005275 * 0.0035 **
Q ₁	Volume flow between adjacent discs.	m ³ /sec	m ³ /sec	m ³ /sec	m ³ /sec
π	Relationship of circle diameter to radius	3.14159	3.14159	3.14159	3.14159
r_1	Inner radius of discs at fluid entry to rotor (m)	0.015	0.015	0.015	0.015
g	Acceleration due to gravity m/s ²	9.81	9.81	9.81	9.81
N	Number of spaces between discs	1	8	4	4

Table 5.2 Explanation of symbols used in the calculations.

A worked example of one of the tests, (test 81), shown below, lists typical results.

$$\text{Efficiency } \eta = \frac{Q\bar{p}_t}{P} = \underline{\underline{6.829 \%}}.$$

$$\text{Reynolds number } N_{RE} = \frac{\rho\Omega h^2}{\mu} = \underline{\underline{624.}}$$

$$\text{Volume rate parameter } U_o = \frac{Q_1}{2\pi\Omega h\bar{r}_1^2} = \underline{\underline{0.0295.}}$$

$$\text{Dimensionless torque } T_1 = \frac{P}{N\phi h\Omega^3 \bar{r}_1^4} = \underline{\underline{2.6}}$$

$$\text{Dimensionless pressure change } p_t = \frac{\bar{p}_t}{\rho\Omega^2 \bar{r}_1^2} = \underline{\underline{0.9766}}$$

These results have similar values to Crawford and Rice, (1974), shown in the list of symbols, except for the Reynolds Number which range from 1 to 10 in the paper. This is low compared to Reynolds numbers in industrial applications which can range from hundreds to several thousands.

Calculations were made to determine the gap between the discs using the Reynolds numbers of 1 to 10 quoted by Crawford and Rice, (1974), and using a speed range of 1000 to 4000 rev/min. The calculated gap ranged from 0.049 to 0.309 mm. How these could be achieved in practice is not understood. Professor Rice had published these results from work at Arizona State University in the 1960's and 1970's. Enquiries of the University indicated that he and his work were unknown and explanation of these figures was not available.

A list of the symbols used with the non dimensional and actual dimensional forms is shown in Table 5.3.

In non dimensional analysis only three symbols are used.

These are:-

Length. This is shown as L. In 'real' units this will be metres, (m)

Mass. This is shown as M. In 'real' units this will be kilograms, (kg).

Time. This is shown as T. In 'real' units this will be seconds, (s).

Symbol	Name	Non dimensional form	Actual dimensional form.
Q	Volume flow rate	L^3 / T	m^3 / s
N	Rotational speed	$1 / T$	$1 / s$
D	Rotor diameter	L	m
g	Acceleration due to gravity	L / T^2	m / s^2
P	Pressure / head	M / L^2	Kg / m^2
ρ	Fluid density	M / L^3	kg / m^3
μ	Viscosity of fluid.	$M / L.T$	$kg / m.s$
P	Power transfer.	$M.L / T$	$Kg.m / s$

Massey, B.S., Mechanics of Fluids, (1995).

Table 5.3, Symbols used for the non-dimensional analysis.

5.4 Description of fluid characteristics.

Fluid behaviour can be divided into Newtonian and non-Newtonian where the difference is in the way the fluid behaves while undergoing a shear force.

A Newtonian fluid is defined as a fluid that has a simple relationship between shear stress and shear rate and the relationship is linear. Water is an example of Newtonian fluid.

Non-Newtonian fluids. The descriptions of Non-Newtonian fluids that follow were taken from Massey, (1995), and Hughes et al., (1967).

Non-Newtonian fluids do not follow a linear stress – strain characteristic. They are in three general classifications.

a) Time independent fluid.

The time independent fluids can be represented by three separate types of fluids and are described as follows.

i) Bingham plastic fluids.

These show a yield stress at zero shear rate which when exceeded follows a straight line relationship between shear stress and shear rate. These fluids tend to be slurries, emulsions and suspensions of solids in liquid.

ii) Pseudoplastic fluids.

This does not show a yield stress but has a progressively decreasing slope of shear stress against shear rate. The slope of this characteristic at any

point is defined as the apparent viscosity at that point. This type of fluid approximates to blood, i.e., red blood cells that are suspended in plasma.

iii) Dilatant fluids.

This type of fluid is similar to a pseudoplastic fluid in that it does not have a yield stress but that the apparent viscosity increases with increasing shear rate. As before, the slope of this characteristic at any point is defined as the apparent viscosity at that point.

Examples of these characteristics are shown in Figure 5.2. This shows that for a Newtonian fluid, the viscosity is a constant value irrespective of the rate of shear as the characteristic between shear stress and shear rate is a straight line.

For a non-Newtonian fluid the viscosity is not a constant value in relation to shear rate, as the characteristic changes slope as the shear rate changes. Figure 5.2 shows two examples, a pseudoplastic fluid and a dilatant fluid.

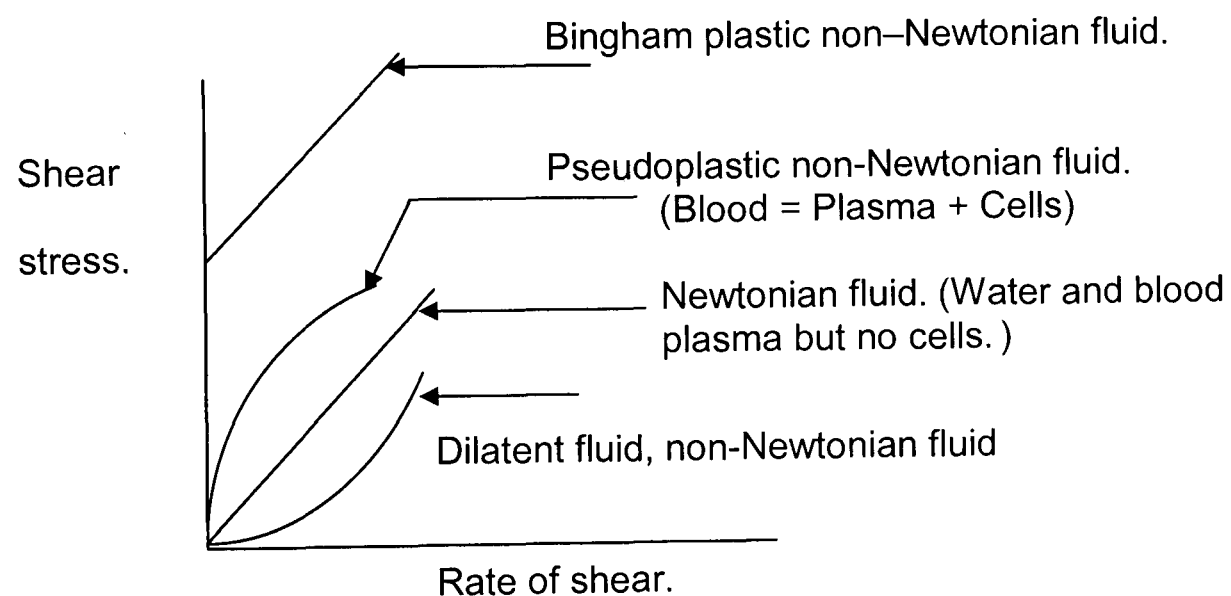


Figure 5.2, Relationships of shear stress and shear rate for fluids.

<http://www.bendigo.latrobe.edu.au/biolsc/phys/appsci/bio30obe/lecture/lecture11.htm>

Massey B.S., (1995)

b) Time dependent fluid.

With this fluid the shear rate is not a single value function of the shear stress but depends on the shear time or the previous shear stress history of the fluid.

There are two basic types of time dependent fluids

i) Thixotropic fluids.

The apparent viscosity decreases depending on the length of time of shearing and the shear rate. As the fluid is sheared the molecular structure breaks down but will reform if the fluid is allowed to rest regaining the original viscosity. An example of a thixotropic fluid would be a printer's ink.

ii) Rheopectic fluids

With this fluid the viscosity increases with shear and the length of time and is the reverse of a thixotropic fluid. The thickening of egg white by beating is an example of this characteristic.

c) Viscoelastic fluids.

The shear strain and shear rate are related to the shear stress and the fluid shows both elastic and viscous characteristics. In a viscoelastic fluid some of the energy of deformation may be recoverable similar to the recovery after deformation of a solid. In a purely viscous fluid all the energy of deformation is lost.

5.5 Test rig construction.

A test rig was built and the layout is shown on figure 5.3.

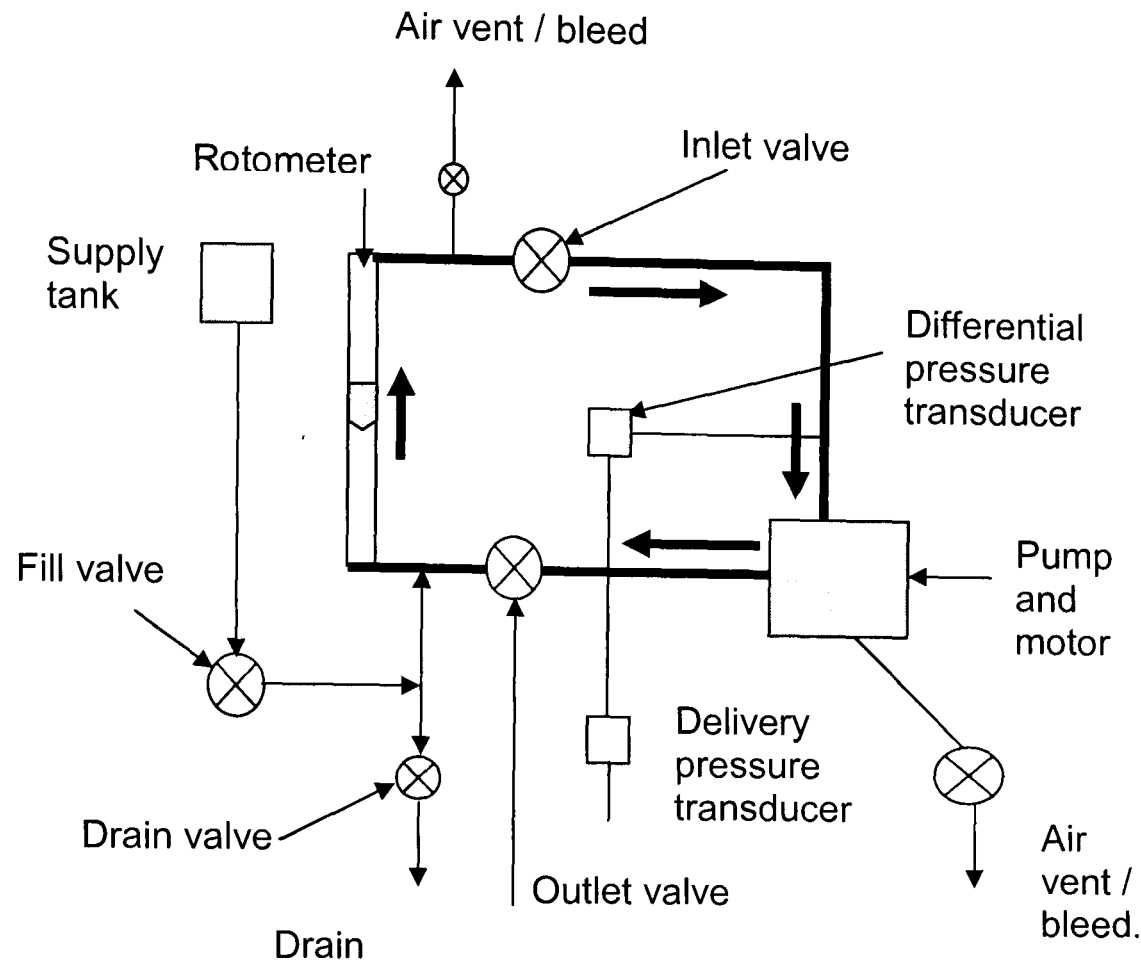


Figure 5.3, Layout of test rig.

The pump was mounted on a 'Dexion' structure that was bolted to a heavy table. The pipe work was made from 25mm, (one inch), bore plastic pipe as used in domestic water systems. Pipe bends were curves rather than 90° corners and the entry length into the pump was a minimum of 10 pipe diameters in order to give a smooth a fluid flow as possible into the pump.

Transducers were used to measure the pressures generated at the fluid inlet and outlet. The readings from the transducers were converted to millimetres of mercury using the formula shown elsewhere in this thesis.

The pump was built from commercially available materials and additional parts such as bearings and seals were purchased as required. The construction of the pump and other details above are covered elsewhere in this thesis.

Power for the motor was supplied from the domestic 240V 50 Hz mains supply, transformed and rectified to 24V DC using a power supply unit. The speed of the 24V DC motor was varied by using a manual control unit. The

motor, power supply unit and control unit, were all purchased from trade suppliers.

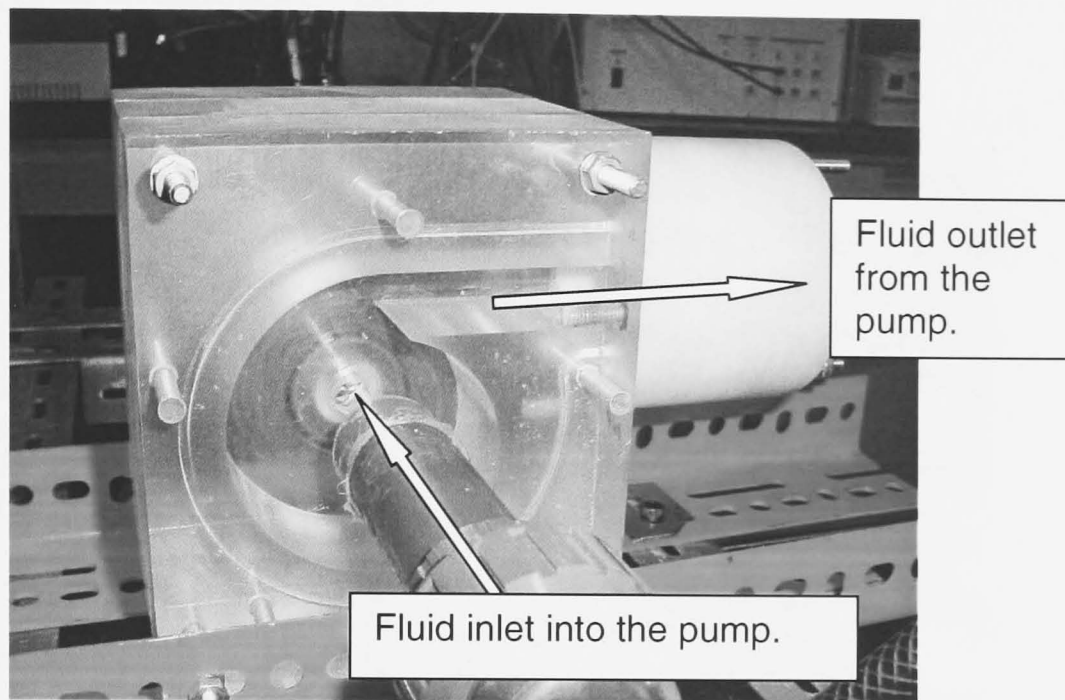


Figure 5.4. The body of the pump.

Figure 5.4 shows the pump casing ready for assembly to the table. The three sections of the pump have been bolted together and the position of the volute and cutwater can be seen above the inlet pipe. The outlet is at the right hand end of the pump. There are no pump internal parts shown in this photograph.

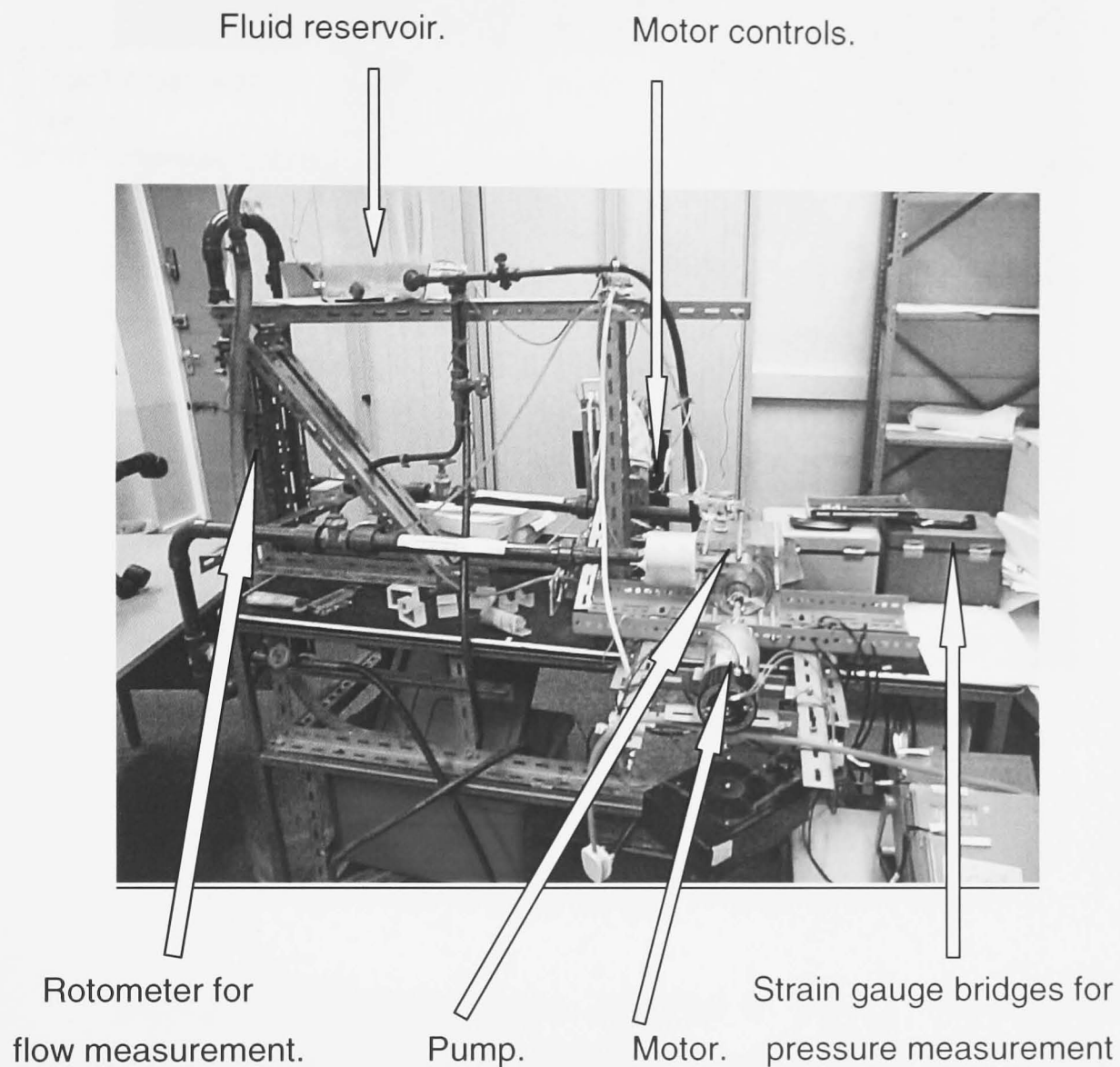


Figure 5.5, General arrangement of test rig.

Figure 5.5 shows the rig from the side. The motor is seen from the tachogenerator end with the pump shaft in line with the motor. The flow measurement device, the Rotometer, is at the extreme left hand end of the rig in this photograph. The pressure measurement and control equipment can also be seen.

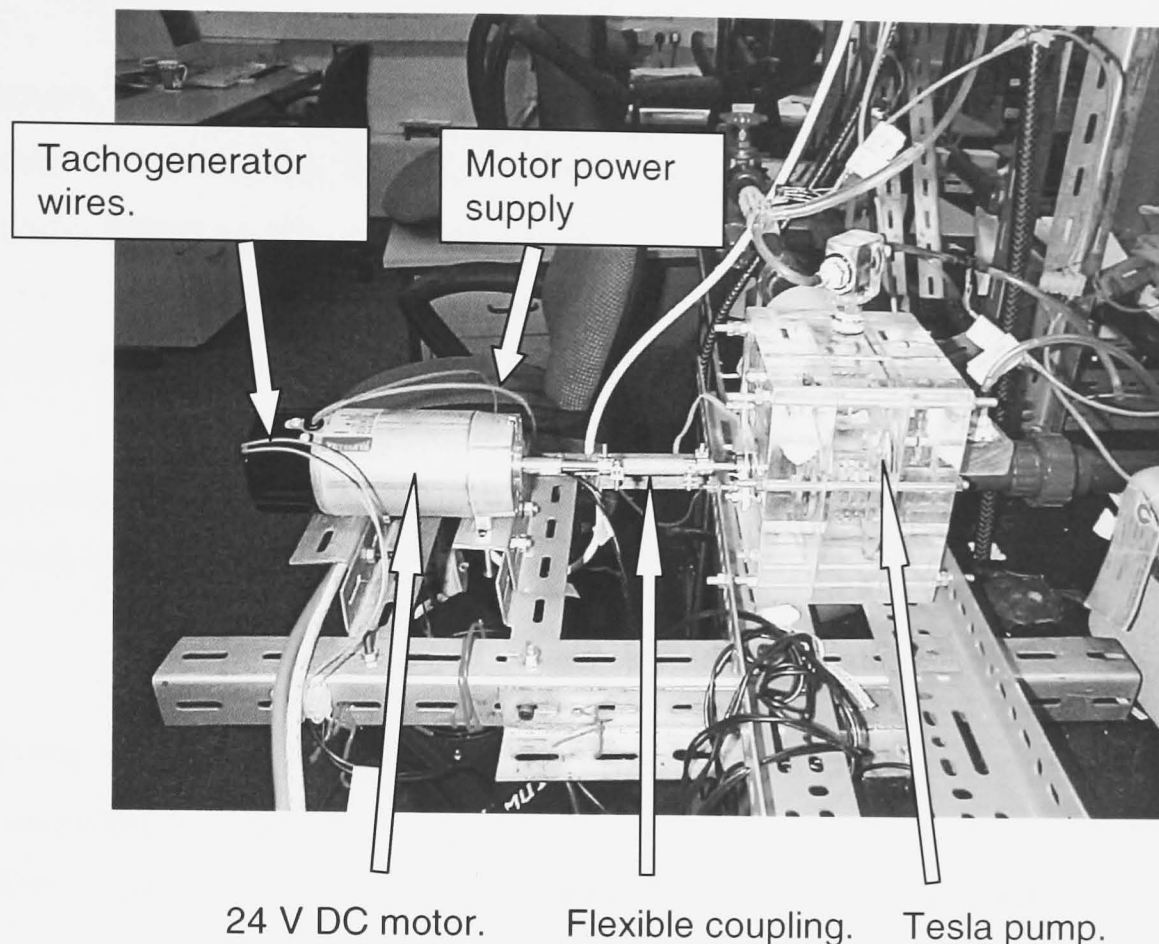


Figure 5.6, General arrangement of motor and pump.

Figure 5.6 shows the pump mounted to the rig. The flexible drive connection from the motor to the pump can be seen. The pump is bolted to a 'Dexion' structure, which was bolted to the table. The two sets of wires coming out of the motor are the motor power supply and the tachogenerator wires.

It was possible to reach pump speeds of over 4000 rev/min. and at this speed pump, deliveries of over 10 l/min were possible at differential pressures of over 250 mm Hg.

The fluid pressures at the pump were calculated using the formula obtained from previous tests measuring strain against pressure.

The rig was fitted with three digital multimeters. Two measured the voltage and current taken by the motor and the third was used to measure the voltage generated by the motor tachogenerator.

The tests measured pump speed (tachogenerator voltage), delivery quantity (l/min), fluid pressures at the pump (strain), motor voltage (volts) and motor current (amperes).

A test programme was written to explore the performances of the three rotors, to decide which was the 'best' rotor and then to concentrate on that rotor to test for the full performance tests. Initial tests were to be conducted using water as the test fluid. Once these tests were completed the test fluid would be changed to a fluid representative of blood in terms of viscosity and any other characteristics that could be obtained.

Each test was run from a 'set point' as long as the pressures did not exceed the rating of the pressure transducers. Two maximum 'set points' were chosen, based on the performance of the human heart. These were

- a) 10 litres / minute at a pressure of 200 mm Hg.
- b) 5 litres / minute at a pressure of 200 mm Hg.

If the pump could achieve this performance, lower pressures and deliveries would be no problem.

The figure for the 'set point' pressure was selected for two reasons,

- a) The maximum blood pressure difference measured in the vascular system can reach 200 mm Hg. This pressure occurs at the outlet from the heart.

- b) The 'set point' pressure at 200 mm Hg is less than the maximum pressure rating of the transducers of 5 lbf/in²., equivalent to 259 mm Hg.

Two sets of tests were taken. The first set explored the characteristics of the pump performance from the 'set point', irrespective of the speed reached. The pump and the valves in the test rig were adjusted to achieve the 'set point', and then changing the pump speed only, recording the performance over the speed range.

The second set of tests was at constant speed. The pump speed and test rig valves were adjusted to achieve the 'set point'. The test rig valve(s) would then be adjusted to change the output of the pump to give a series of output deliveries, ensuring that the pump speed was as constant as possible at each setting. Most tests were to be taken by closing the pump outlet valve with the inlet valve wide open. This would give the test fluid an unrestricted entry to the pump but a restricted outlet.

A small number of tests were to be taken with the outlet valve fully open and the inlet valve adjusted to give the target 'set point'. This would give

an unrestricted outlet from the pump but with a restricted inlet. This was intended to determine if cavitation was possible.

5.6, Results.

5.6.1 Test conditions

The 'set points' chosen for the first set of tests are shown in Table 5.4.

<u>Delivery l/min</u>	<u>Pressure difference mm Hg</u>
10	200
10	150
10	100
5	200
5	150
5	100

Table 5.4. The performance 'set points' for the initial tests.

The differential pressure is the pressure difference between the pump inlet and outlet. With this connection, the pressure transducer senses the inlet and outlet pressures and records the difference between them but not the pressures relative to atmosphere or absolute. This method gives a measure of the pump pressure performance without reference to external pressures.

Another set of performance 'set points' was chosen for the constant speed tests and these are shown in Appendix 12.

Initial tests were conducted to ensure that the pump operated properly and to fix any leaks and other problems. These tests were not part of the test sequence and were not recorded.

5.6.2 Test fluids, introduction.

In order to achieve satisfactory results it was considered necessary to conduct tests to find a fluid that was representative of blood. Enquiries from Wickramasinghe R. of Colorado State University, by e-mail dated July 2003, indicated that there was not an 'industry standard' fluid used by all

researchers. This statement was surprising considering the importance of the work and the number of papers published on the subject of blood pumps and blood contacting devices. Many researchers appear to use a mixture of water-glycerine or water-xanthan gum. Checks indicated that there were no health and safety problems with these mixtures.

The pump tests were to be conducted using three different test fluids. These are described below.

The first fluid used was tap water. This was allowed to settle in order to ensure that there were no air bubbles. Before each test the pump was run to ensure that there was no air entrained in the system. The water was not treated or mixed with any other fluid.

The second fluid was water – glycerine.

A third set of tests used a mixture of water - xanthan gum. Preparation of these fluids is described later.

Thurston, (1996), discusses analog fluids for human blood and indicates that a safe, stable, reliable alternative has not been found but that some fluids can be used to imitate blood in selected ways. A mixture of 45% glycerol in water gives a viscosity that is close to that of blood at 25°C. It is low cost and transparent but does not give the viscoelastic flow properties of blood.

Wickramasinghe, (2002), showed results using a mixture of 45% glycerine to 55% water by weight. This gives a fluid that is similar to blood in terms of viscosity but does not have the shear - viscosity relationship of blood.

Brookshier and Tarbell, (1993), indicate that mixtures of water / xanthan gum / glycerine can do a '*reasonably good job of modelling the flow of blood in large arteries*'. They do not recommend a preferred specification for a test fluid but instead state that the xanthan gum / glycerine solutions '*should be useful blood analog fluids for in vitro hemodynamic studies*'.

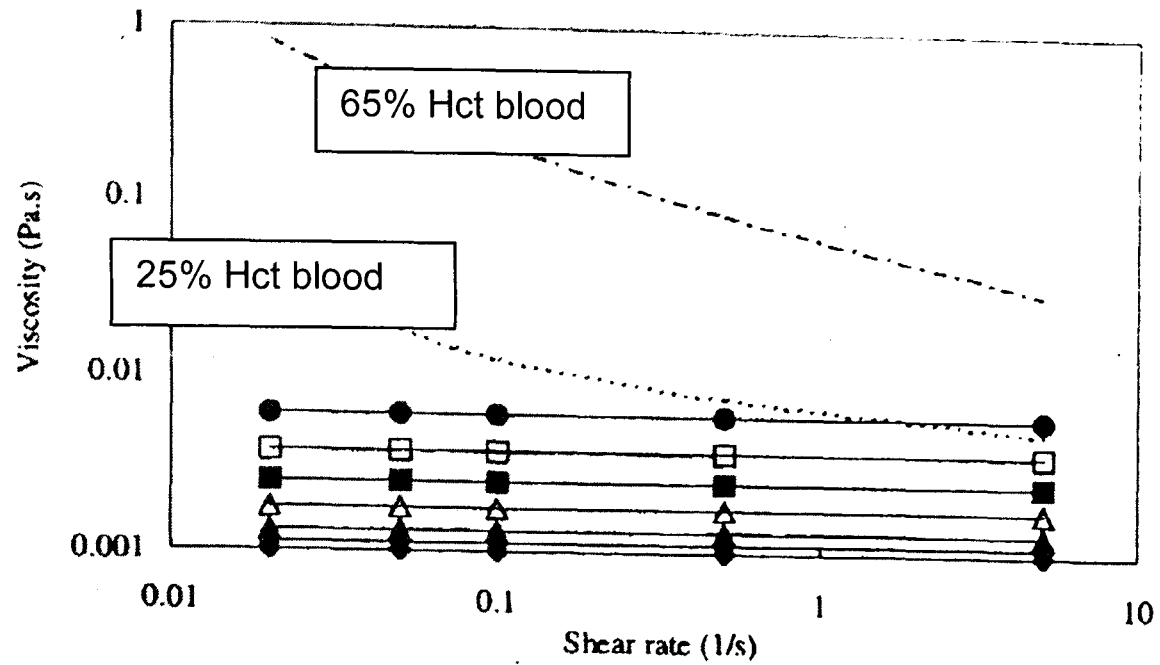
These mixtures seemed to be the best available at the time and accordingly several small samples of different mixture strengths by weight were prepared.

Although this was used as a test fluid, a water - glycerine mixture it is not equivalent to blood in all respects. Figure 5.7 shows the results of tests by Wickramasinghe, (2002), to determine the shear and viscosity relationship of

water and glycerol mixtures and comparing them to blood at hematocrits of 25% and 65%. (Normal blood hematocrit is around 45%). These tests show that the viscosity for each mixture did not change relative to shear rate. This is the characteristic of a Newtonian fluid in which the flow and shear rate are always proportional to the applied stress and where the viscosity depends on temperature. It is obvious that Newtonian fluids do not accurately model blood over a useful range of shear rates

Assuming that the normal blood hematocrit is 45%, none of the water / glycerine mixtures were suitable to simulate blood at any of the shear rates shown. Other mixtures of water - glycerine could have been made and tested but it is assumed that the water - glycerine characteristic would meet the slope of the blood characteristic at only one value of shear rate where the two lines would cross.

Taking tests with many water - glycerine mixtures, it would be possible to obtain a result that conformed exactly to the shear / viscosity relationship of blood at any one hematocrit but it would be of little practical use over a range of shear rates.



From Wickramasinghe R., (2003)

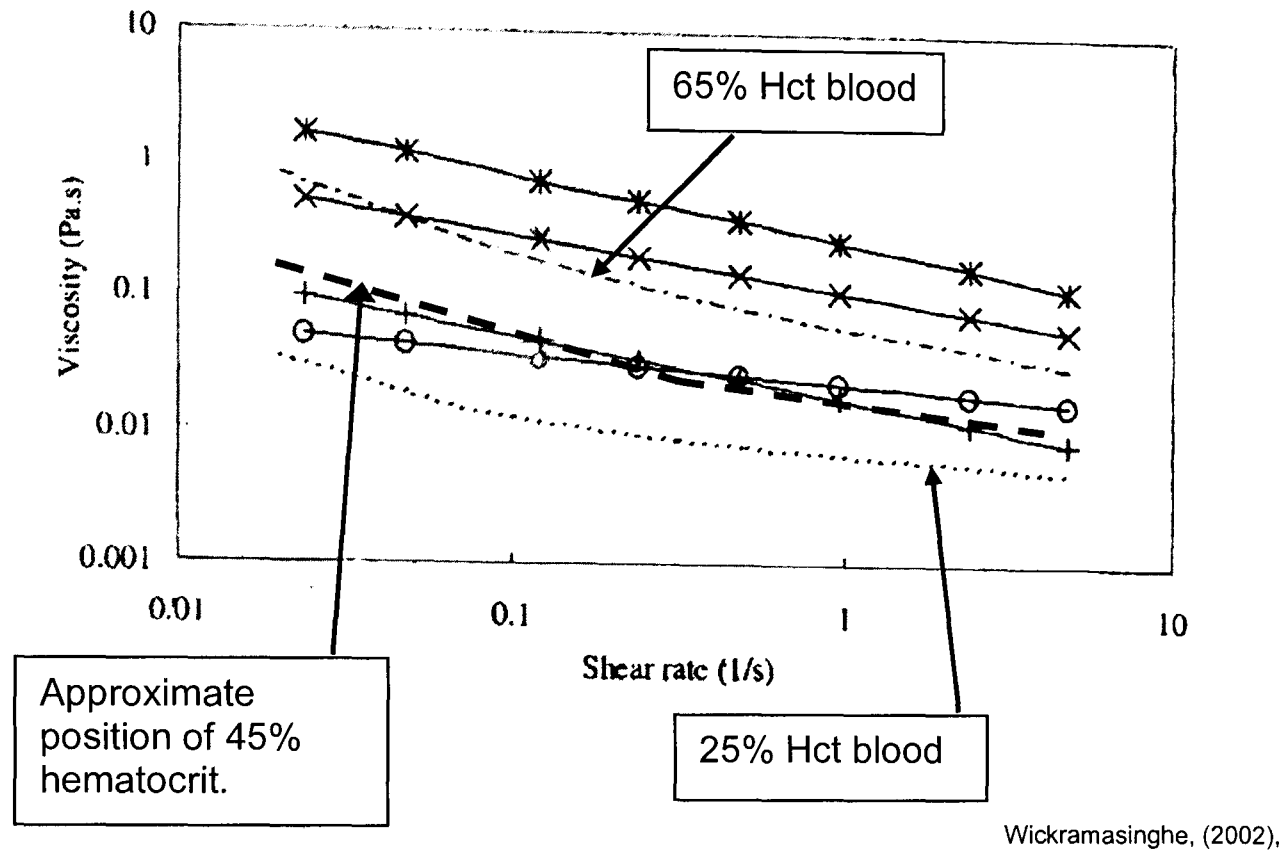
Fluid	Water %	Glycerol %	Symbol
A	100	0	◆
B	95	5	◇
C	90	10	▲
D	80	20	△
E	70	30	■
F	60	40	□
G	50	50	●

From Wickramasinghe R., (2002)

Figure 5.7, Viscosity and Shear rate of water / glycerine mixtures compared to blood.

Other results published by Wickramasinghe, (2002), Figure 5.8, use test fluids made from mixtures of water / glycerine / xanthan gum. The results indicate that these are shear thinning fluids where the viscosity is lower when a greater shear rate is applied.

This is the characteristic of a non-Newtonian fluid, where the flow and shear rate are not in proportion to the applied stress and the viscosity changes with the shear rate. This characteristic is similar to blood.



Fluid	Water %	Glycerol %	Xanthan gum%	Symbol
H	59.9925	40	0.0075	○
I	59.96	40	0.04	×
J	99.99	0	0.01	+
K	49.925	50	0.075	⌘

From Wickramasinghe R., (2002)

Figure 5.8 Variation of viscosity with shear rate for various non Newtonian fluids.

Using the lines for 65% hematocrit blood and 25% hematocrit blood as guides, the approximate position of 45% hematocrit blood is superimposed on the figure.

Assuming that the position of the blood hematocrit line of 45% is correct, the results indicate that the water - xanthan gum mixture, fluid J, has the closest relationship to 45% hematocrit blood than any of the other

mixtures as it has almost the same slope and crosses the 45% hematocrit line.

Test fluid preparation.

Preparation of water – glycerine mixture.

Tests were conducted to determine the specific gravity and the percentage of glycerine in samples of water - glycerine mixtures. The specific gravity of water was measured as 1.000 and the specific gravity of 100% glycerine was measured as 1.250. The figure for the specific gravity for laboratory grade glycerine taken from the catalogue of 'Fisher Scientific' products is 1.257 – 1.261.

From these two values of specific gravity for water and 100 % glycerine and assuming that the relationship between the specific gravity and the glycerine content was a straight line, the calculated result for 45% glycerine should have been a specific gravity of 1.1125. However tests conducted at various mixtures did not lie on the straight line connecting these two results.

The results were plotted and using a 'best' fit curve, gave a curve with a quadratic equation of $y = 0.000006x^2 + 0.001945x + 1.000117$ as shown on Appendix 13.1. Based on this plot the specific gravity for a mixture of 45% glycerine in water was 1.099792.

Re-plotting the same results using a 'best' straight-line plot gave a line with an equation of $y = 0.0025x + 0.9899$ as shown on Appendix 13.2. Based on this plot the specific gravity for 45% glycerine was 1.1024.

Tests on a sample gave a specific gravity of 1.100 for 45.7 % glycerine. The Handbook of Chemistry and Physics, (1946), quotes a specific gravity of 1.1138 for a 45% aqueous glycerol solution with a viscosity of 4.509 @ 20° C. Although not stated it is assumed that the viscosity is quoted in centipoises.

For the same water / glycerine mixture, Lönn and Wulff, et al., (1997), quoted the density as 1,113 kg/m³ at 21deg C and the dynamic viscosity as 3.4*10⁻³ kg/m.s. The dynamic viscosity of blood, with a red blood cell volume of 40%, was quoted as 3.5*10⁻³ kg/m.s. Kaye and Laby, (1995), quoted the density of blood as 1,055 Kg/m³.

Considering the above results, the average figure for the specific gravity of a 45% glycerine / water mixture is 1.1069 and the chosen figure for specific gravity of 1.100 for the water - glycerine mixture was considered accurate enough for test purposes. It could be read easily from a standard hydrometer and was repeatable. Any difference in the results due to variations in specific gravity of the fluid around this figure that could not be read using a standard hydrometer were considered to be insignificant

Preparation of water – xanthan gum mixture.

Following Wickramasinghe, (2002), samples of a water- xanthan gum mixture were made. The mixture selected was 99.99% water, 0.01% xanthan gum. This was the simplest mixture shown and from the results on Figure 5.8, appeared to give a reasonably good relationship with 45% hematocrit blood when viscosity was plotted against shear.

The addition of the gum was intended to give a non linear relationship between shear rate and viscosity and to simulate the performance of blood under shear forces where blood is considered to be a shear thinning Non-Newtonian fluid.

Four bottles of the water - xanthan gum mixture were made by filling each bottle with 5 litres of water and then adding 0.5 gram of xanthan gum powder.

Some difficulty was found when mixing the water and gum.

For bottle no. 1, it was noted that if the quantity of gum, 0.5 gram, was dropped directly into 5 litres of cold water, the gum tended to form small and large clumps of something like a glue. If a clump were broken up, dry powder could be seen in the centre of the clump. Shaking the bottle did not improve the mixing and clumps could be seen floating around in the bottle.

For bottle no. 2, 0.5 gram of powder was added to 100 ml of cold water. This was then placed on to a laboratory mixer and heater table where the water was heated and stirred until it almost boiled. This produced a mixture that appeared to be homogenous throughout and was a better result than before. This was then added to cold water where some clumps could be seen but were less in number than before.

For bottle no. 3, 0.5 gram of powder was added to 100 ml of water which was then placed into an ultra sonic cleaner. The action of the cleaner was not effective in breaking up the clumps of powder and was discontinued after 10 minutes. This sample was heated and stirred as before. This gave a result similar to bottle no. 2.

For bottle no. 4, 150 ml of water was heated but without boiling. 0.5 gram of powder was sprinkled on to the surface of the hot water as slowly as possible while the water was stirred. The powder was not allowed to 'clump' on the surface of the water. As stirring progressed the powder appeared to mix better than before. This gave the best result of the tests conducted, with no noticeable clumps floating around in the mixture. As this powder is used as a cooking additive it may be possible that heat is necessary to make it react and mix properly.

A hydrometer was used to measure the specific gravity of water and one of the water / gum mixtures. The specific gravity of water had a value of 1.00 as expected. The specific gravity of the xanthan gum / water mixture no. 1 was measured at 0.998 which was less than the figure for water. The test was repeated with the same result.

5.6.3 Viscosity tests.

Tests were taken to determine the viscosity of the fluids. A small test rig was constructed comprising upper and lower reservoirs, a length of capillary tubing, a pressure transducer and pipe work such that the pressure difference across the length of the capillary tube could be measured while the fluid was flowing. The tube length was measured as 502 mm and the bore was calculated as 0.583 mm. The pressure transducer had been previously calibrated. Chapter 5.1, Pressure Measurement, details the procedure.

The transducer calibration constant was

$$0.02883 \text{ Strain} \equiv 1 \text{ mm Hg}$$

A measuring cylinder was used to collect the fluid that passed through the tubing and the quantity collected was timed. The height of the fluid reservoir compared to the centre of the capillary tube could be varied by either moving the tube or the reservoir. The change of fluid level in the reservoir

during each test was considered to be negligible. The general arrangement is shown in Figure 5.9.

A minimum of 4 tests was taken at each height setting and the results were averaged to give a point for plotting.

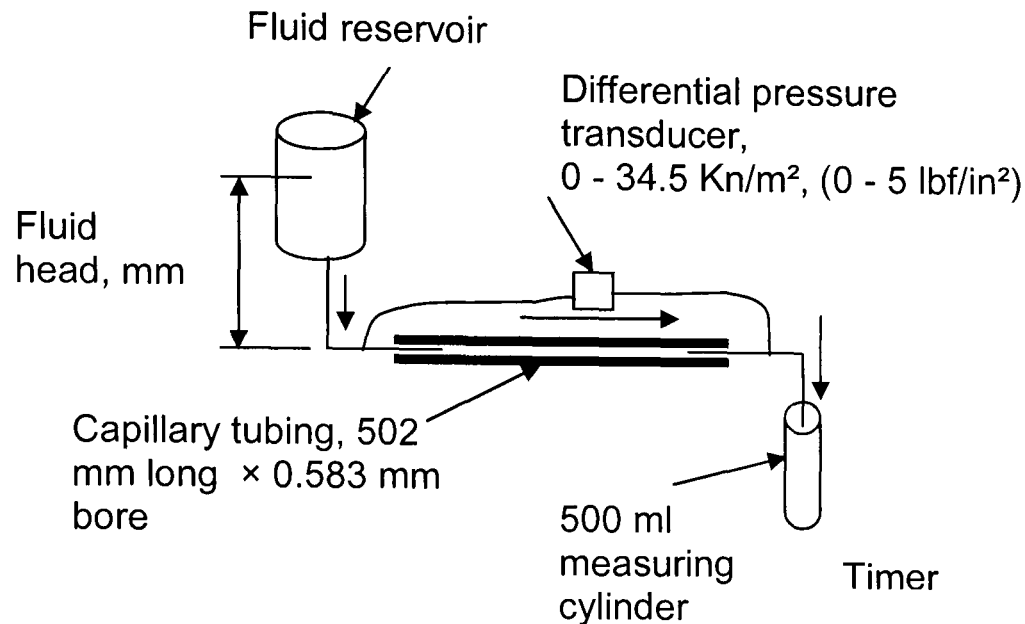


Figure 5.9 Schematic arrangement of viscosity test rig.

5.6.4 Results of viscosity-shear tests.

Tests were taken to confirm,

- a) the viscosity of water in order to prove the accuracy of the method used.
- b) the viscosity – shear rate relationship of a water – glycerine mixture
- c) the viscosity – shear rate relationship of a water – xanthan gum mixture.

Water only test.

The results of the water only test are shown in Figure 5.10. The average of the readings gave a viscosity for water of 0.00096731. The viscosity from the plot was 0.0009356011. These figures were close to the quoted viscosity of 0.001002 @ 20 ° C from Rogers and Mayhew, (1983).

The slope of the result, 0.0000000485, indicated that the viscosity of water varied very little with shear and was considered to be not shear sensitive and therefore a Newtonian fluid.

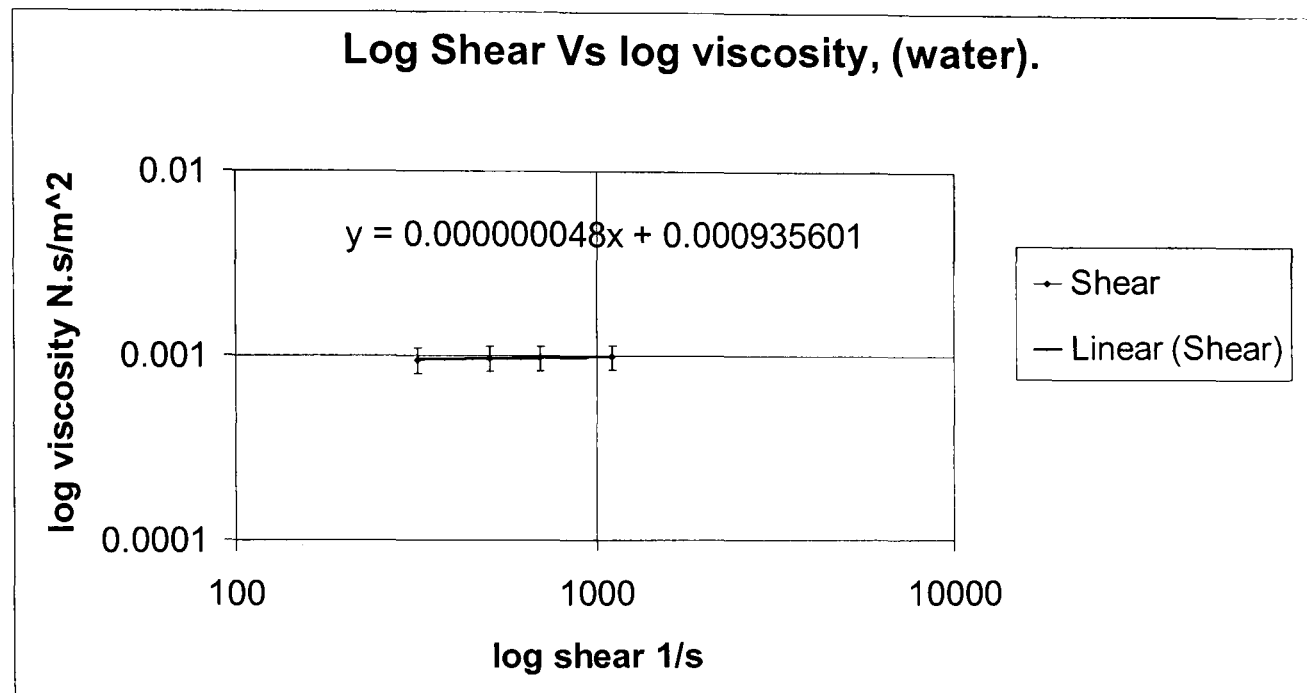


Figure 5.10 Result of the viscosity / shear relationship of water.

It was not possible to compare this result with Figure 5.7, from Wickramasinghe, (2002). Although the viscosity result was almost identical at 0.001, the maximum shear rate for Wickramasinghe, (2002) was below 10 where the Middlesex University shear was between 100 and 1000.

Water / glycerine test.

These tests were conducted using the same test rig and method of calculation as the water only tests.

The results are shown in Figure 5.11. Taking the average of the readings gave a viscosity of 0.0042262. Plotting the results and fitting a 'best' line indicated that the viscosity of the water / glycerine mixture is around 0.0037406. The 'flatness' of the slope at 0.00000258 indicates that this mixture is not shear sensitive and is a Newtonian fluid.

It was not possible to compare this result with Figure 5.8 directly. Although the viscosity was between the results for 50% glycerine and 40%

glycerine, the maximum shear rate for Wickramasinghe, (2002), was below 10 where the Middlesex University shear was between 100 and 1000.

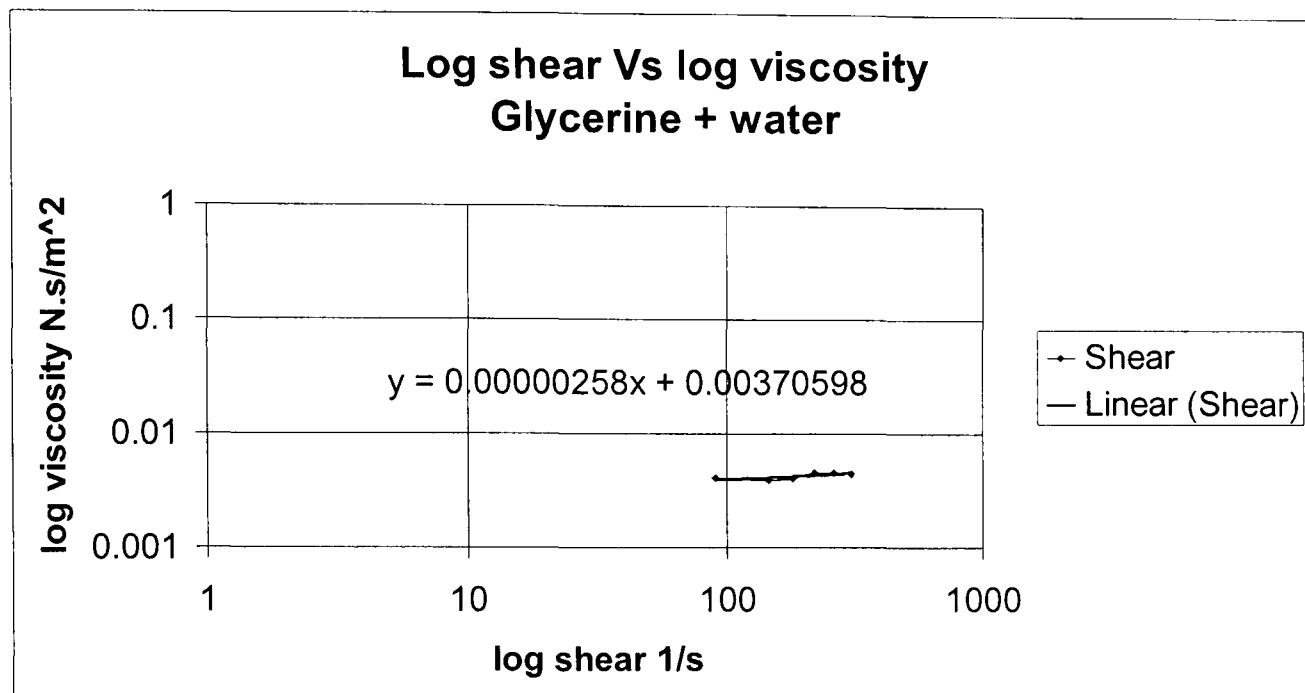


Figure 5.11 Results showing the viscosity / shear relationship of the water glycerine mixture.

The water – glycerine viscosity at 0.0042262 is close to the quoted viscosity of a 50% mixture quoted at 0.00420 @ 30° C by Kaye and Laby, (1995). Taking the results from Wickramasinghe, (2002), and estimating a viscosity for 55% water / 45% glycerol mixture gave a viscosity of 0.00494. Thurston, (1996), indicates a Newtonian viscosity of 0.00397 @ 25° C for a 45% mixture.

Water / xanthan gum test.

Tests were conducted using the same test rig and method of calculation as the previous tests using a water / xanthan mixture in the same proportions as quoted by Wickramasinghe, (2002).

The results are shown in Figure 5.12 and indicate that the effective viscosity of the water / xanthan gum mixture is around 0.0011 but the slope at 0.00000001 does not clearly indicate that this mixture is shear sensitive and it is difficult to determine if it is a Newtonian or non Newtonian fluid.

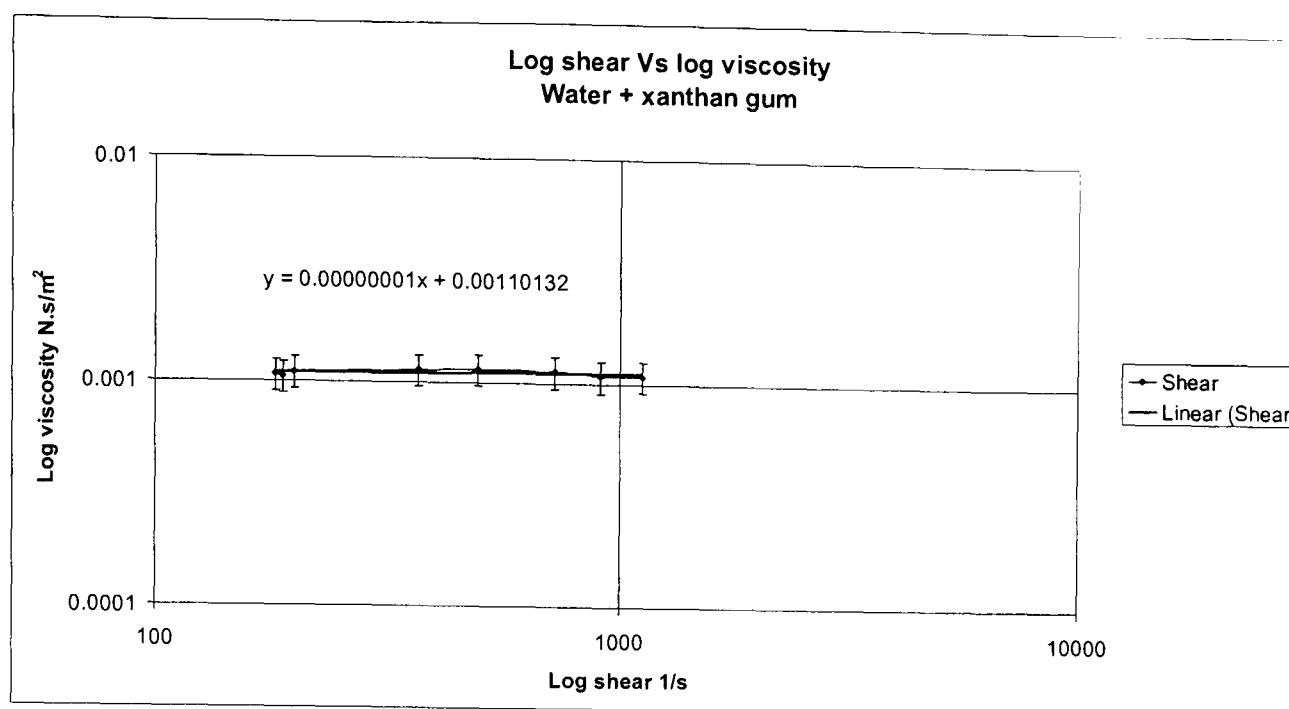


Figure 5.12 Results of the viscosity / shear relationship of the water xanthan gum mixture.

The results were compared to normal haematocrit blood and the results by Wickramasinghe, (2002). They are not comparable with the Wickramasinghe results as it was not possible to compare the shear values quoted by Wickramasinghe.

Klabunde, (2002), indicated that blood viscosity was 3 to 4 times that of water. The viscosity of water can be found and multiplying by 3.5 would give an easy answer for the viscosity at 0.003507 but would not take account of the change of viscosity of blood due to temperature and other conditions.

It is not understood how shear values in the range 1 to 10 were obtained by Wickramasinghe, (2002). With the water – xanthan gum tests, the smallest fluid ‘head’ used was about 3 mm and still gave a shear value of above 100.

Thurston, (1996), described the use of xanthan gum as a blood analogue but indicted that a solution in the ratio of 500 ppm of gum in distilled water, (0.05%), gave a viscosity below that of blood. He considered that a mixture of xanthan gum and glycerine could be made that would be a better blood analogue than either of the two fluids when used separately.

The amount of xanthan gum quoted by Thurston, (1996), is 5 times the amount quoted by Wickramasinghe, (2002) and reinforces the impression that

the mixture tested was effectively water and would explain the poor results of these tests.

5.6.5 Pump tests.

These tests are intended to show the performance of the pump and are in two parts. The first part was to determine what appeared to be the best rotor using water as a test fluid. The second part was to re-test the pump with the 'best' rotor fitted using the alternative test fluids.

The rotor builds and test fluids are shown in Table 5.5 where each rotor and fluid combination are referred to by the rotor identification number.

No. of discs	Fluid	Rotor identification
0	Water	0
3	Water	3
3	Water plus glycerine	3G
3	Water plus xanthan gum	3X
3	Calculated result for blood.	3B
7	Water	7

Table 5.5, Rotor identification.

Once the initial tests had been completed, and the 'optimum' rotor had been selected, additional tests were taken with the optimum rotor fitted but with changes to the test fluid.

In the second part of the tests, the first test fluid was a mixture of water and glycerine. The second test fluid was a mixture of water and xanthan gum. Chapter 5.6.3 describes how the viscosity – shear results for the test fluids were determined. An additional calculated result for blood was added. This used the viscosity of blood as 0.0035 N.s/m^2 taken from Lönn and Wulff, et al., (1997).

An additional set of calculations was made to determine the non dimensional performance of the pump. The formulae for these calculations were taken from Crawford and Rice, (1974). The results of these calculations

will provide values that can be used for comparison with other pumps. This procedure is detailed in Chapter 5.3.1, Non dimensional coefficients.

It was not possible to complete the full series of tests. The pump developed a serious leak and it was not possible to repair the pump and finish the tests before a campus move took place. The results for comparison 7a with rotor 3X were therefore never taken. Additional tests for a test fluid comprising a water / glycerine / xanthan gum mixture were also not taken.

5.6.6 Results.

Introduction.

The first section will describe the determination of the 'optimum' rotor build based on a simple comparison of rotor speed against delivery using water as a test fluid. The rotor giving the required delivery at the lowest speed was considered to be the 'optimum' rotor for each of the performance 'set points' as listed in Table 5.6. Tests would also be taken with no restrictions to flow by setting both inlet and outlet valves fully open. The resultant speeds were expected to be lower than the other tests.

Other tests were to show if performance changes could be detected if the output was controlled by restricting the inlet with the outlet unrestricted. These tests were also used to examine the flow to observe if any bubbles could be seen as this would indicate cavitation.

The second set of tests describes the pump performance with the optimum rotor fitted to the pump and using the various test fluids.

The performance with each rotor configuration is shown in a series of constant speed tests. These use the 'set point' figures as a starting point for each test to provide a comparison with each rotor build. The pressures used at the 'set points' will be the pump differential pressure which is a better measure of pump performance than the delivery pressure. It provides a comparison with other pump performances taken from published papers.

Comparison number.	Performance 'set point'
1a	10 l/min @ 200 mm Hg.
2a	10 l/min @ 150 mm Hg
3a	10 l/min @ 100 mm Hg
4a	5 l/min @ 200 mm Hg
5a	5 l/min @ 150 mm Hg
6a	5 l/min @ 100 mm Hg
7a	10 l/min @ both valves open
8a	10 l/min @ 200 mm Hg, inlet valve set / outlet valve wide open
9a	5 l/min @ 100 mm Hg, inlet valve set / outlet valve wide open

Table 5.6. Performance 'set points' for all the pump tests.

Determination of the optimum rotor.

The pump was tested with water only and with the three rotor builds 0, 3, and 7 as listed in Table 5.6.

From the results on Table 5.7, the results at the 'set point' only, all three rotors were able to deliver the required set point performance but that there are noticeable differences in speed between the three rotors. Considering an output of 10 l/min @ 200 mm Hg, rotor 3 gave the lowest speed at 3492 rev/min while rotor 0 gave the worst performance at 4003 rev/min. The performance of rotor 7 was similar to rotor 3. These results indicated that the rotor build effected pump performance.

The results from the first series of tests are shown in appendices 14.1 to 14.9. These show that in practically every test, rotor 0 needed a higher speed to generate the required flow than either of the other two rotors.

The results with rotors 3 and 7 were similar to each other and it was difficult at times to determine the 'better' of the two from the results. It eventually was decided to use rotor 3 for the following reasons.

a) The results between rotors 3 and 7 were very similar but rotor 3 appeared to have the advantage in most tests.

b) The gaps between the discs of rotor 3 are greater than with rotor 7. If used with blood, the reduced number of discs and smaller surface area of rotor 3 would give less chance of blood clots forming. The performance with this rotor would therefore be more representative of the 'real' application.

The tests with the restricted inlet and open outlet valve settings, Appendices 14.8 and 14.9, did not show any signs of cavitation. Comparison of the speed Vs delivery performance at 10 l/min @ 200mm Hg with inlet valve open and inlet valve restricted did not show any appreciable difference between the two results as shown on Appendix 14.10. A similar result was obtained for 5 l/min @ 100 mm Hg. as shown on Appendix 14.11.

Following the completion of the above tests, the optimum rotor, rotor three, was fitted to the pump. A set of constant speed tests were then taken to determine the constant speed characteristics of the pump. Each speed was set to the required figure using the motor controls and the delivery was set using the outlet valve with the inlet valve wide open. Results were taken at speeds of 1000 to 4000 rev/min in 500 rev/min steps and the pump output 'set points' were the same as for the rotor optimisation tests. Frequent checks were made to ensure that the speed remained constant during the test.

The results of the tests are shown in Appendix 14.12 to Appendix 14.17

Appendix 14.12 shows the flow Vs delivery pressure of the pump at the setting point of 5 l/min. The result shows that for each flow Vs delivery pressure curve, the pressure is relatively constant over the flow range for the speed selected. The pressures generated were approximately 270 mm Hg at 4000 rev/min and 10 mm Hg at 1000 rev/min.

Appendix 14.13 shows the flow Vs differential pressure of the pump. The results show the same characteristics as the previous result, Appendix 14.12. The pressures at the setting point of 5 l/min were 270 mm Hg at 4000 rev/min and 20 mm Hg at 1000 rev/min.

Comparison no.	Performance 'set point'	Rotor 0, rev/min	Rotor 7, rev/min	Rotor 3, rev/min	Rotor 3G, rev/min.	Rotor 3X, rev/min
1a	10 l/min @ 200 mm Hg	4003	3635	3499	3177	3443
2a	10 l/min @ 150 mm Hg	3464	2986	2959	2708	2905
3a	10 l/min @ 100 mm Hg	2833	2538	2414	2143	2330
4a	5 l/min @ 200 mm Hg	4016	3500	3445	3149	3367
5a	5 l/min @ 150 mm Hg	3386	2989	3020	2658	2884
6a	5 l/min @ 100 mm Hg	2773	2478	2373	2148	2306
7a	10 l/min @ both valves open	1245	1078	1003	950	0
8a	10 l/min @ 200 mm Hg, inlet valve set / outlet valve wide open	3534	3565	3455	3152	3395
9a	5 l/min @ 100 mm Hg, inlet valve set / outlet valve wide open	2768	2390	2289	2133	2308

Table 5.7. Comparative results for Flow Vs Speed with the three rotors and three fluids at the quoted 'set point'

The power used by the pump was also recorded. The results on Appendix 14.14, show that the power is relatively constant over the flow range for the speeds selected. The power taken at the 5 l/min setting point was 120W at 4000 rev/min and 5W at 1000 rev/min.

The above tests were repeated at a setting point of 10 l/min with very similar characteristic results as detailed above.

Appendix 14.15 shows the flow Vs delivery pressure at the 10 l/min setting point. The results show a similar characteristic to the 5 l/min results. The pressures at the setting point of 10 l/min were 275 mm Hg at 4000 rev/min and 10 mm Hg at 1000 rev/min.

Appendix 14.16 shows the flow Vs differential pressure at the 10 l/min setting point. The results are similar to the 5 l/min results. The pressures at the setting point of 10 l/min were 275 mm Hg at 4000 rev/min and 20 mm Hg at 1000 rev/min.

Appendix 14.17 shows the flow Vs power at the 10 l/min setting point. The results are similar to the 5 l/min results. The power taken at the 10 l/min setting point was 120W at 4000 rev/min and 5W at 1000 rev/min.

The relative 'flatness' of the performance curves are not the usual constant speed performance characteristics of a pump. These points will be discussed later in this thesis.

Results with the alternative test fluids.

Discussion of results with the optimum rotor fitted.

A second set of tests was conducted with the optimum rotor fitted but using the test fluids simulating blood from the hydraulic point of view.

These fluids are the water - glycerine and the water – xanthan gum mixtures described previously in Chapter 5.6.2.

Figure 5.13 indicates that there appears to be a 'best' rotor in terms of minimum speed for the target performance. Rotor 3G gives the lowest speed for any set performance compared to the other rotor builds at the same performance setting. There is also a 'grouping' of the performance from these results. Examination of the results on Table 5.7 and Figure 5.13 indicate that the set point pressure appears to be the 'link' between these results.

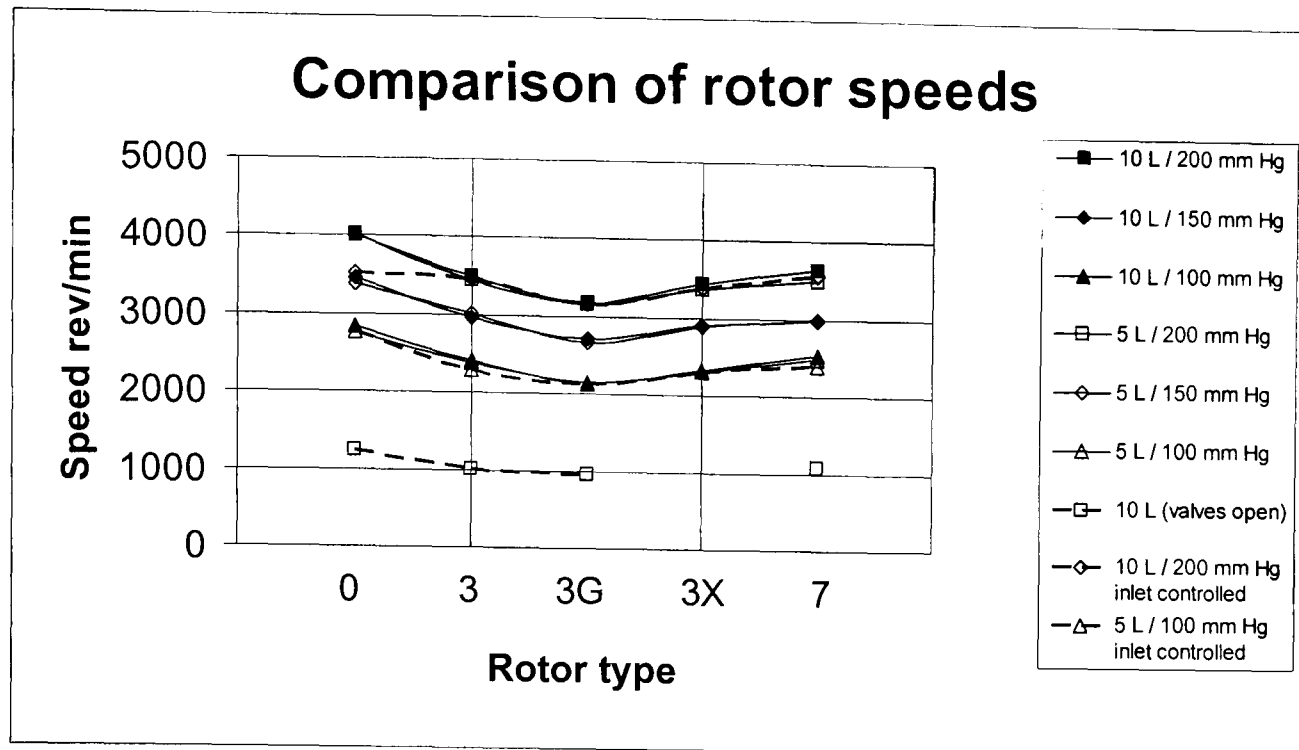


Figure 5.13. The relationship between all the rotor types and speed at the 'set points' for all the test fluids used.

From Figure 5.13, at the same differential pressure of 200 mm Hg., the speeds obtained for the three results at 10 l/min., 5 l/min and 10 l/min with the inlet controlled, are almost identical for each rotor build and test fluid used. The max and min rotor speeds for this pressure were 3499 and 3445 rev/min respectively, (1.5% difference).

The same relationship of speed and pressure is seen at a pressure of 150 mm Hg, (0.98% difference), and at a pressure of 100 mm Hg, (5.46% difference). This result implies that the pressure is a function of rotor speed. This in turn indicates that any other pump of the same type, but of a different rotor diameter, should achieve the same pressure if the rotor is running at the same peripheral speed.

A three dimensional plot of the full set of comparisons at the 10 l/min @ 200 mm Hg 'set point' is shown on Figure 5.14. These results include all the fluids and all the rotor configurations. This shows that rotor 0 needed the highest speed to meet the target performance. Rotor 3G gave the lowest speed to meet the target performance of all the rotors and fluids tested. This fluid had the highest viscosity of all the fluids tested.

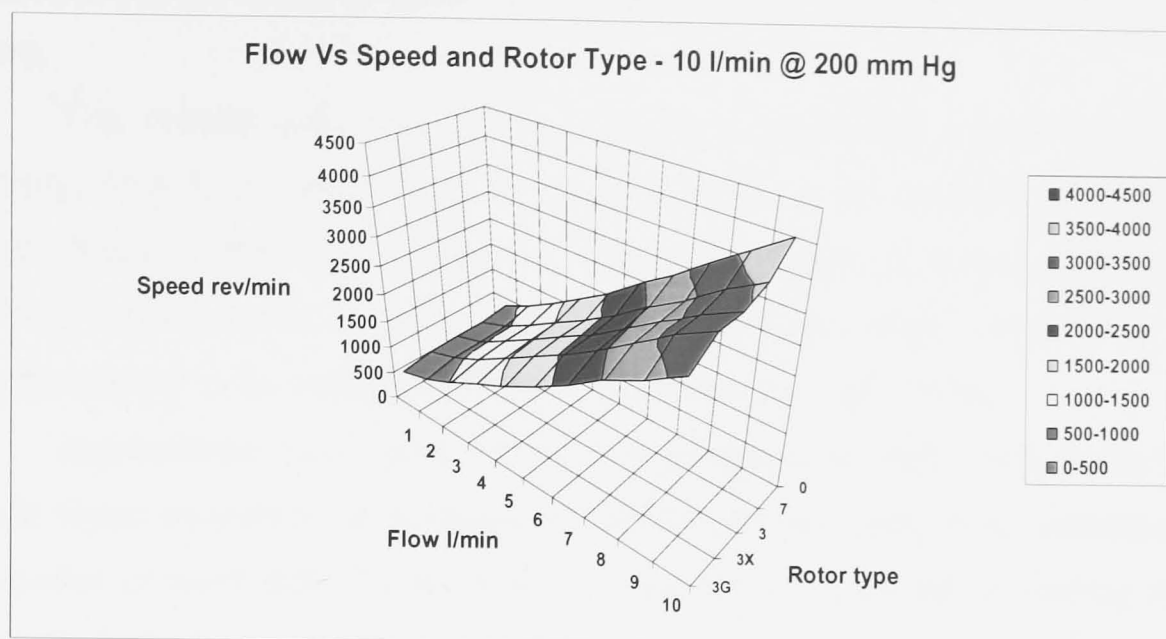


Figure 5.14. Three dimensional representation showing the relationship between speed, flow, rotor type and test fluid at the setting point of 10 l/min @ 200 mm Hg.

Discussion of Flow Vs Speed tests

The results in Appendix 14.18 show the performance of the three rotors 0, 3 and 7 using water and the alternative test fluids at a 'set point performance' of 10 l/min @ 200 mm Hg. The 'best' performance was considered to be the lowest speed for the target 'set point'.

From Appendix 14.18, rotor 3G gave a better performance compared to rotor 3 as the result showed the lowest speed for the same delivery. This is the performance difference between water and water / glycerine. Taking the figures from Table 5.7, there is a speed difference of 315 rev/min. Rotor 3X resulted in a lower speed, 49 rev/min, than rotor 3 for the same delivery, as shown in Table 5.7. This was the difference in speed between water and water / xanthan gum. From these results, it was considered that the rotor could be optimised relative to any given test fluid and the controlling characteristic of the fluid could be viscosity as there was a greater change in viscosity between the water and water / glycerine mixture compared to the water and water / xanthan gum mixture. The results from the full series of tests are shown in Appendices 14.18, 14.25, 14.32, 14.39, 14.46, 14.53, 14.60, 14.67 and 14.74. These show that in every test except one, rotor 0

needed a higher speed to generate the required flow than either of the other rotors.

The results also show rotor 3G gave the lowest speed for the set delivery. This fluid was nearest to blood in terms of the viscosity of the fluids tested. It was difficult to distinguish between the results of rotors, 3, 3X and 7, as they tended to be superimposed on top of each other. In general these results tended to lie between the results of the other two rotors.

Appendices 14.67 and 14.74 were intended to determine if the pump would show cavitation as during these tests the fluid entry was restricted and the outlet unrestricted. There were no indications of cavitation during any of these tests and the speeds were close for the same deliveries and pressures as shown by a comparison of Appendices 14.18 and 14.67 for 10 l/min @ 200 mm Hg and Appendices 14.53 and 14.74 for the 5 l/min @ 100 mm Hg. Appendices 14.19 to 14.24, with the outlet valve controlling the delivery of 10 l/min @ 200 mm Hg, are repeated in Appendices 14.68 to 14.73 but with the inlet valve controlling the delivery. These tests are repeated at 5 l/min @ 100 mm Hg comparing Appendices 14.54 to 14.59, (outlet valve control), and Appendices 14.75 to 14.80, (inlet valve control). There is little or no difference in each set of comparative results and there was no evidence of cavitation in any of the tests. This indicates that the pump performance can be controlled by outlet or inlet restriction with little or no effect on the performance.

From the result for Appendix 14.60, 10 l/min. 'as found', the speeds are the lowest recorded for all the tests. This test was taken with both the inlet and outlet valves fully open and gave the lowest pump speed for the target delivery. All the results were replotted as a three dimensional surface plot and are shown in Figure 5.15. This shows the results at the setting points for all the rotors and test fluid combinations used. The result for 7a is slightly distorted as one result was not possible. This is shown on Table 5.7 as zero.

Figure 5.15 shows that rotor 3G gives the lowest speed of any of the rotors for the same comparison number as shown by the shaded bands, indicating the speeds, running across the plot. The comparisons with the lowest outputs, 3a and 6a, (10 l/min and 5 l/min at 100 mm Hg), give lower speeds than those comparisons with the higher outputs, 1a and 4a, (10 l/min and 5 l/min at 200 mm Hg), for any rotor build. The lowest speed of all the

tests was with comparison 7a, when both the inlet and outlet valves were fully open and the pump was not producing any pressurised flow. Comparing the results for rotors 3, (water only), 3X, (water / xanthan gum) and 3G, (water / glycerine), rotor 3G gives the lowest speed. This indicates that fluid viscosity affects pump performance and that the higher the fluid viscosity the better the pump performance.

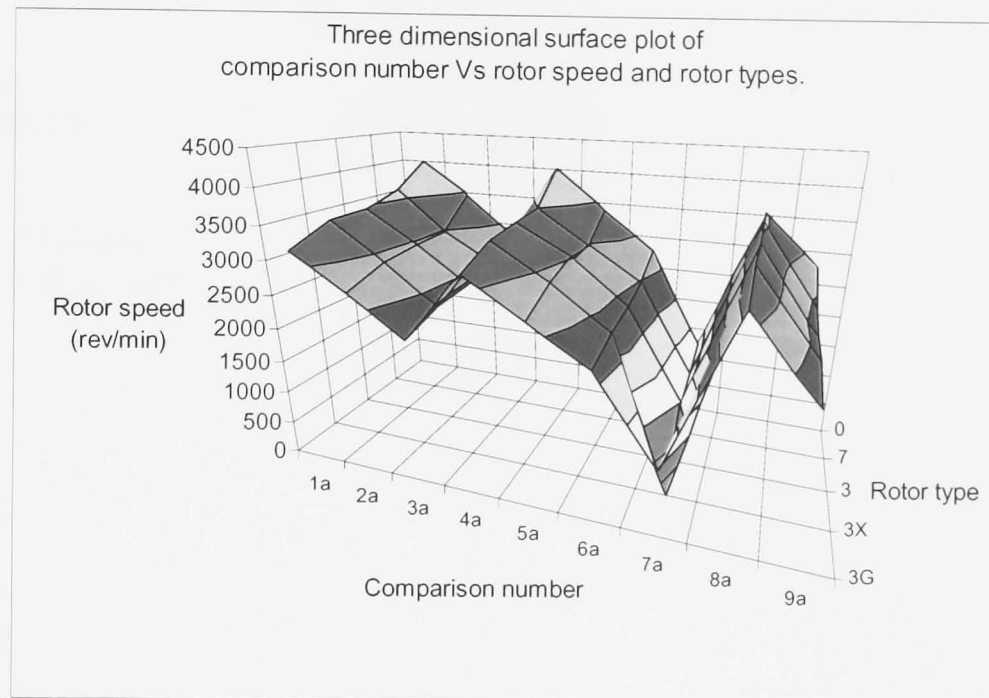


Figure 5.15. Speed comparison of all rotors with all test fluids shown as a three dimensional surface plot.

Blood shear stress calculation.

Now that some pump speeds have been determined, an estimation of blood stresses can be made. Following the theory outlined in Chapter 1.3, Blood, and using the maximum speed found from the test results, a calculation was made of the maximum shear stress likely to be experienced by the blood with this pump. The position inside the pump that was likely to give the highest shear stress in the absence of CFD results was in the gap between the face of the first disc at the maximum diameter and the stationary wall of the volute as shown in Figure 5.16.

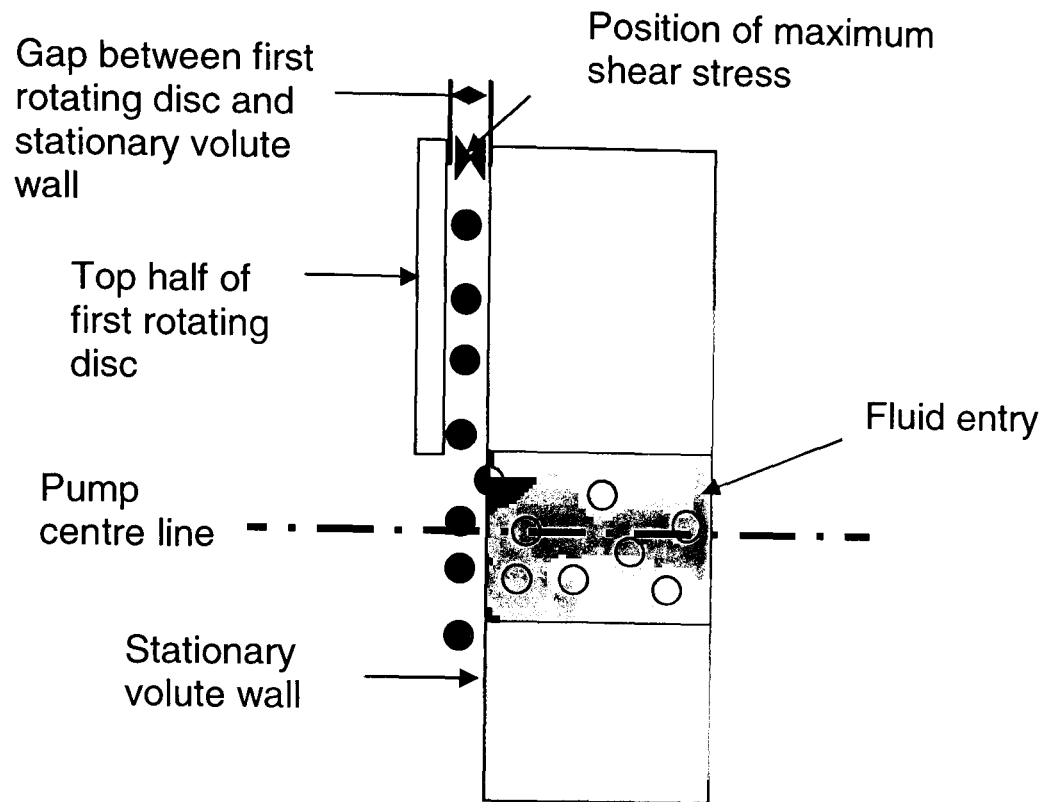


Figure 5.16 Relative positions of the first disc of the rotor and the volute wall.

The stress calculations used the following figures from the pump build dimensions: -

Rotor diameter, 60 mm, the gap between the first disc of the rotor and the volute wall, 1.5 mm and pump speed, 4000 rev/min. The fluid was assumed to be blood with a viscosity of 0.0035 Kg/m.s. A calculation using these conditions gave a shear stress of 29 N/m².

This is less than the figures quoted for red cell damage of approximately 200 N/m² following the results of Sutura and Mehrjardi, (1975) as shown in Chapter 1. This is above the figure of approximately 15 N/m² indicated for damage to the platelets quoted in Chapter 1.3.6. Limiting the pump speed to give a blood shear stress of 15 N/m² gives a pump speed of approximately 2050 rev/min under these build conditions. Figure 5.17 shows the result of the calculations.

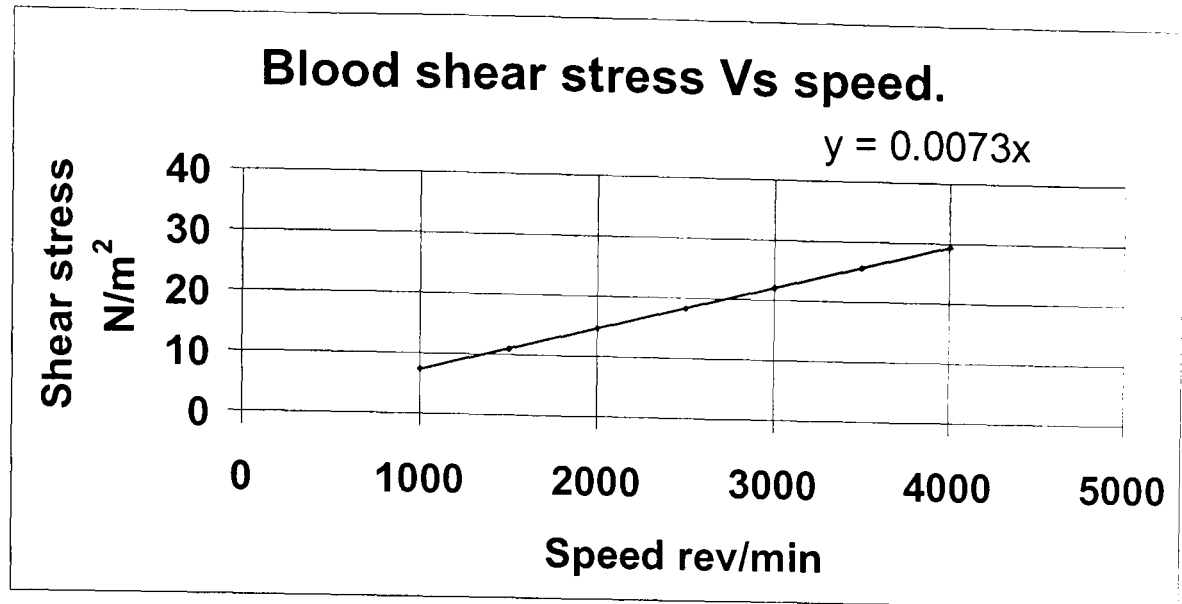


Figure 5.17, Calculated blood shear stress against speed for the pump build tested.

Figure 5.18 shows the speeds and gaps for a pump built to produce a given shear stress at the periphery of a 60 mm diameter disc. This result shows that at a speed of 4000 rev/min for a stress of 10 N/m², the gap will be about 4.5 mm, but at the same speed for a stress of 40 N/m², the gap reduces to about 1 mm.

Following these calculations, the build tolerances of blood pumps will need careful consideration if shear stress problems are to be avoided. Additional calculations concerning the gap and stress values will be made later in this thesis.

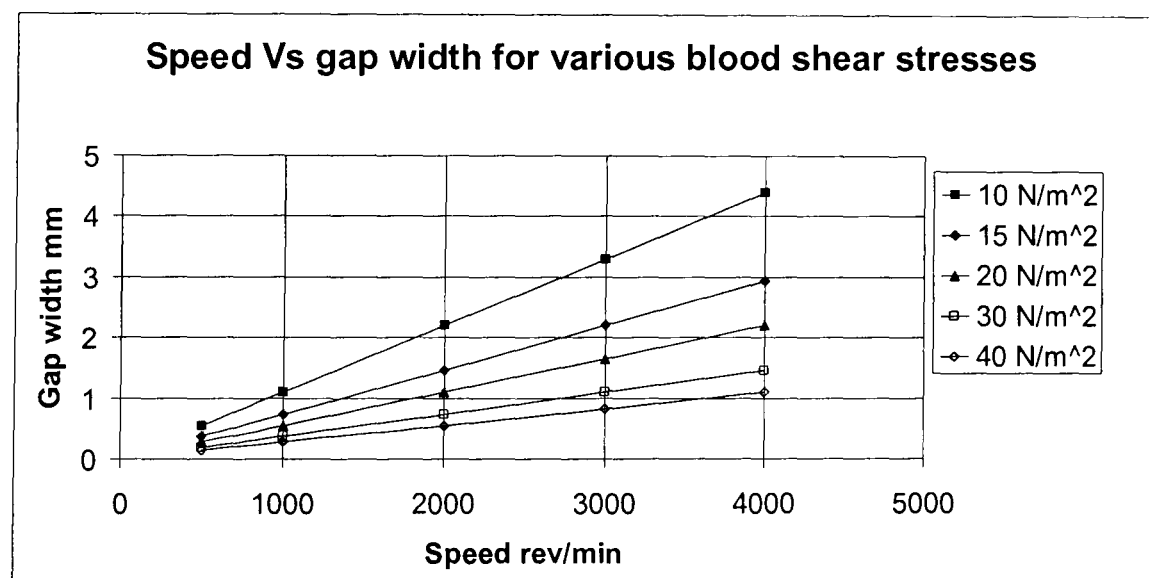


Figure 5.18 The calculated gap against speed for given blood shear stresses.

Discussion of Flow Vs Pressure tests

The three dimensional representations of delivery pressure and differential pressure at a set point of 10 l/min @ 200 mm Hg are shown on Figures 5.19 and 5.20. They indicate that the pressures generated are not effected by the rotor type and fluid combination, as for a given flow rate the pressure appears to be the same across the rotor type axis as indicated by the shading bands running across the plot. These plots do not indicate pump speed.

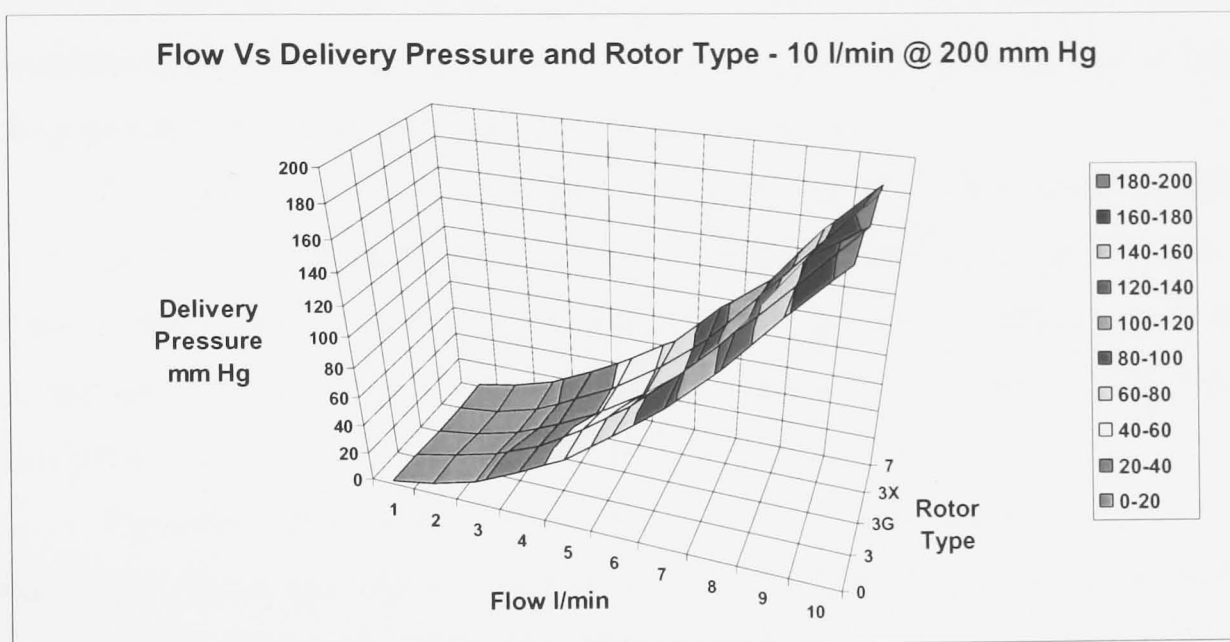


Figure 5.19. Three dimensional representation of delivery pressure Vs flow and rotor type.

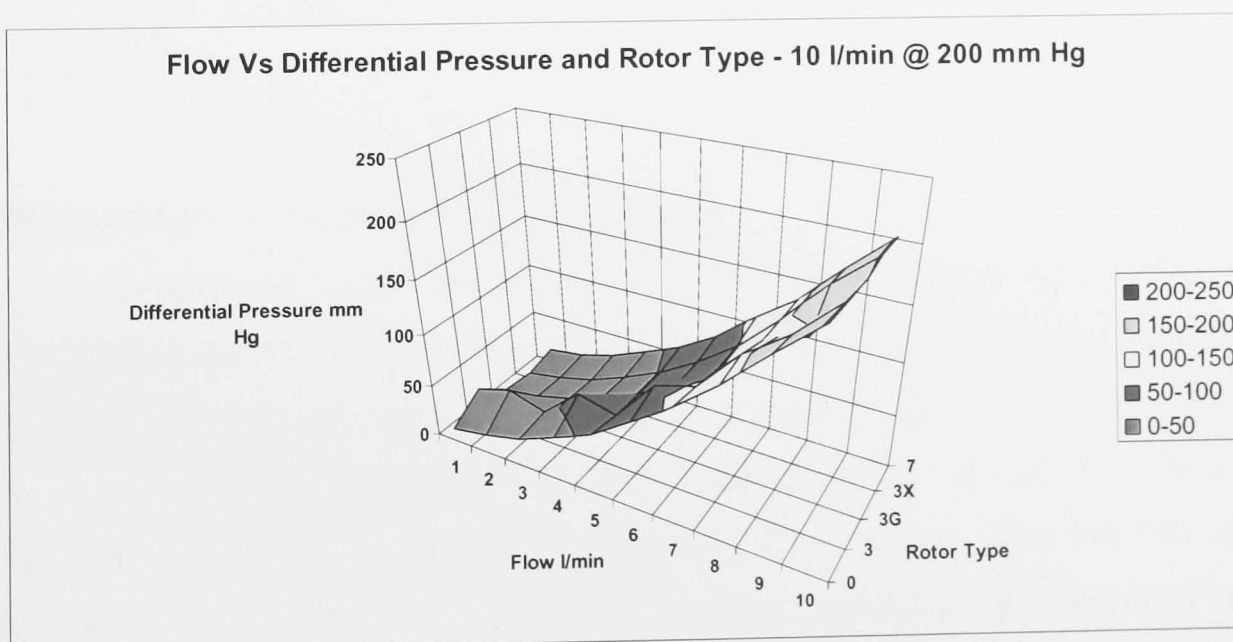


Figure 5.20. Three dimensional representation of differential pressure Vs flow and rotor type

Appendices 14.81 to 14.86 show the pump performance at constant speed for delivery pressure and differential pressure at each of the setting points. The motor was set to give the required speed and the valves were set to give the required flow. The motor speed was checked at every reading to ensure accuracy.

The results show that the pressure is essentially constant over the delivery range tested. This reinforces the impression that speed is the controlling factor determining the pressure generated by this pump type. Assuming identical rotor peripheral speeds, a 'small' pump running 'fast' may generate the same pressures as a 'large' pump running 'slow' but a 'large' pump would have a greater delivery at that pressure.

A comparison of the delivery pressures can be seen on Appendices 14.15, 14.81 and 14.85. These are the constant speed results at 10 l/min for all three fluids used. The results for 3500 rev/min show the delivery pressures are higher with the water / glycerine fluid at approximately 240 – 250 mm Hg than either of the other two fluids at 190 – 210 mm Hg.

The differential pressure results, shown on Appendices 14.16, 14.82 and 14.86, show the same characteristic at 3500 rev/min with a pressure of approximately 250 mm Hg for the water / glycerine fluid and 215 to 200 mm Hg for the other two fluids.

Lower speeds show the same pattern but with lower values as expected.

Discussion of Flow Vs Power tests.

Appendices 14.87 and 14.88 show the performance of the pump at constant speed for power at each of the 10 l/min setting points previously listed using the water / glycerine and water / xanthan gum fluids. In each test, the motor was set to give the required speed while the valves controlling the flow rate were set to give the required flow. The power shown is the value required to pump the fluid and is not the power required to run the pump itself. The motor speed was checked at every reading to ensure accuracy.

The results show that the power characteristics are essentially constant with little change over the delivery range tested.

With some of the results it is noted that there is a difference in power consumption at the same speed depending on the fluid used.

The results are plotted on two graphs to make the results clearer.

From Appendix 14.87, (1000 to 2500 rev/min), at each speed, the water / glycerine fluid takes the most power while the water / xanthan gum fluid takes the least power. The lowest power consumption of 10 Watts, or less, occurs at speeds of 1500 rev/min or below. A similar result is shown in Appendix 14.88 (2500 to 4000 rev/min), where the results at 3500 rev/min show that the power requirement is greater with the water / glycerine fluid, at approximately 90 – 95 W, than with the other two fluids at 62 – 80 W.

Appendix 14.89 shows the power requirement at 5 l/min for water and for water / glycerine. At 3500 rev/min the water requires 70 W while the water / glycerine fluid requires approximately 100 W.

Lower speeds show the same pattern as before but with lower values as expected.

The data for water / xanthan gum fluid at 5 l/min was not taken due to problems with the pump. The tests with the water / glycerine mixture at 4000 rev/min were not undertaken as the pressures exceeded the maximum rated pressure quoted by the manufacturer of the pressure transducers.

Discussion of Flow Vs Efficiency results.

From the results on Table 5.8, the 'set point' only, rotor 3G shows the highest efficiency and rotor 0 shows the lowest efficiency. The overall values of the efficiencies are low, the highest is less than 11%. The results for rotor 3B are not test results and should be ignored

From Figure 5.21, the efficiency increases as the pressure and delivery combination reduces. This indicates that the efficiency increases as the pump load reduces. Pump load is understood to be a combination of delivery and pressure.

Comparison no.	Performance 'set point'.	Rotor 0, %.	Rotor 7, %	Rotor3, %	Rotor 3G, %	Rotor 3B, %	Rotor 3X, %
1a	10 l/min @ 200 mm Hg	4.113	4.996	5.965	6.829	6.829	6.767
2a	10 l/min @ 150 mm Hg	4.798	7.595	7.076	8.102	8.102	8.366
3a	10 l/min @ 100 mm Hg	5.854	7.268	8.623	10.532	10.532	10.499
4a	5 l/min @ 200 mm Hg	2.047	3.251	2.861	3.263	3.263	3.885
5a	5 l/min @ 150 mm Hg	2.828	4.025	3.327	3.974	3.974	4.569
6a	5 l/min @ 100 mm Hg	3.299	4.239	4.289	5.464	5.464	5.978
7a	10 l/min @ both valves open.	6.398	8.190	9.331	10.743	10.743	-
8a	10 l/min @ 200 mm Hg, inlet valve set / outlet valve wide open.	4.003	6.092	5.825	6.193	6.193	7.431
9a	5 l/min @ 100 mm Hg, inlet valve set / outlet valve wide open.	3.245	4.358	4.849	4.379	4.379	5.674

Table 5.8, The comparative results for Flow Vs Efficiency with the three rotors and three fluids at the quoted 'set points'.



Figure 5.21. Relationship between all the rotor types and efficiency at the 'set points' for all the test fluids used.

The comparisons of results at 10 l/min @ 200 mm Hg and 5 l/min @ 100 mm Hg show that the performances are almost identical. It would appear that there is little or no change in efficiency if the pump output is controlled by throttling the inlet or the outlet.

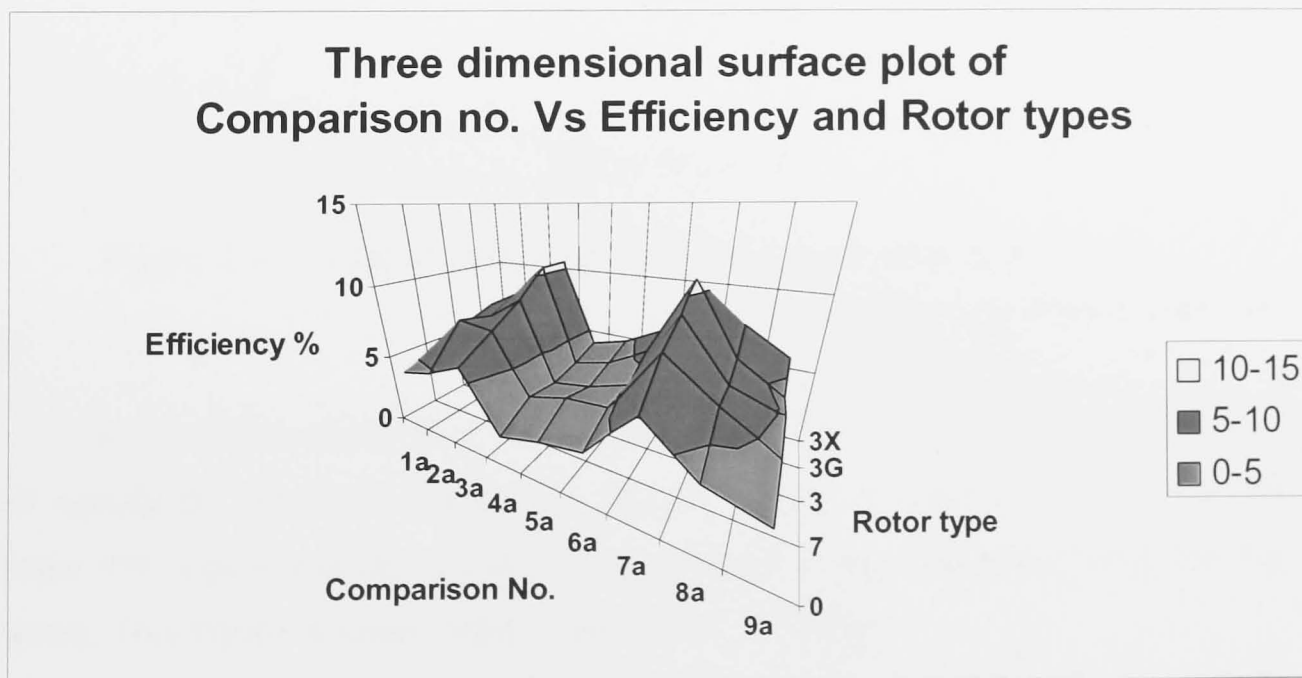


Figure 5.22. Three dimensional representation of Efficiency Vs Flow and rotor type at the 'set points'.

Figure 5.22 is a three dimensional representation of efficiency Vs flow and rotor type at the 'set points'. It shows that the efficiency varies with rotor build and pump performance. The maximum efficiency is around 10% which is low when compared to other pumps where maximum efficiencies of 80% + can be expected.

The reasons for the low efficiency are unknown but it could be that the pump is not an ideal design and construction. The disc diameter ratios at 2 to 1 may be one reason. It is known that small pumps are not as efficient as large pumps as friction and other factors are in larger in proportion. For example, it is not possible to make clearances and surface finishes proportionally comparable between a large pump and a small pump and seals and bearings take proportionally more power to drive with a small pump. Lazarkiewicz and Troskolański, (1965), indicate that one of the disadvantages of impeller pumps is low efficiency at small discharges and high pressures. This type of pump could have similar characteristics.

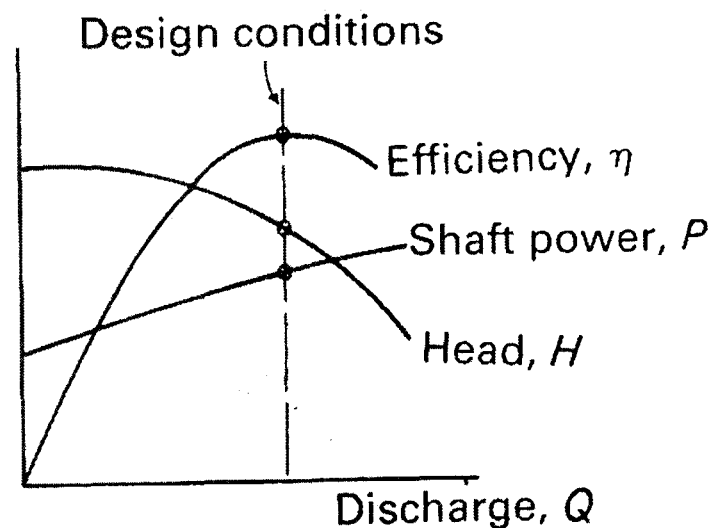


Figure 5.23, Typical performance curves for a centrifugal pump.

B.S. Massey, Mechanics of Fluids, 6th edition. (1989). p549

It should be noted that the shapes of the plotted curves from the Tesla test results do not meet the typical shapes shown in Figure 5.23, where this shows the typical performance of a centrifugal pump with backward sloping blades. This figure is taken from Massey B.S., (1989).

Two examples of the low efficiency results are shown in Figures 5.24 and 5.25 as three dimensional representations of efficiency at setting points of 10 l/min @ 200 mm hg and 5 l/min @ 200 mm Hg respectively. These results

indicate that the efficiency is low at low fluid flow conditions but it increases as the flow rate increases. The efficiency reaches a maximum of around 7% to 8% at 3-4 l/min for the 10 l/min setting point and 4% to 5% at 2-3 l/min for the 5 l/min setting point.

The effect of rotor type and test fluid can also be seen. In the results, the increase in efficiency between rotor 0 and rotor 3 confirms the original decision to use rotor 3 as the optimum rotor and shows that pump efficiency improves as the fluid viscosity increases.

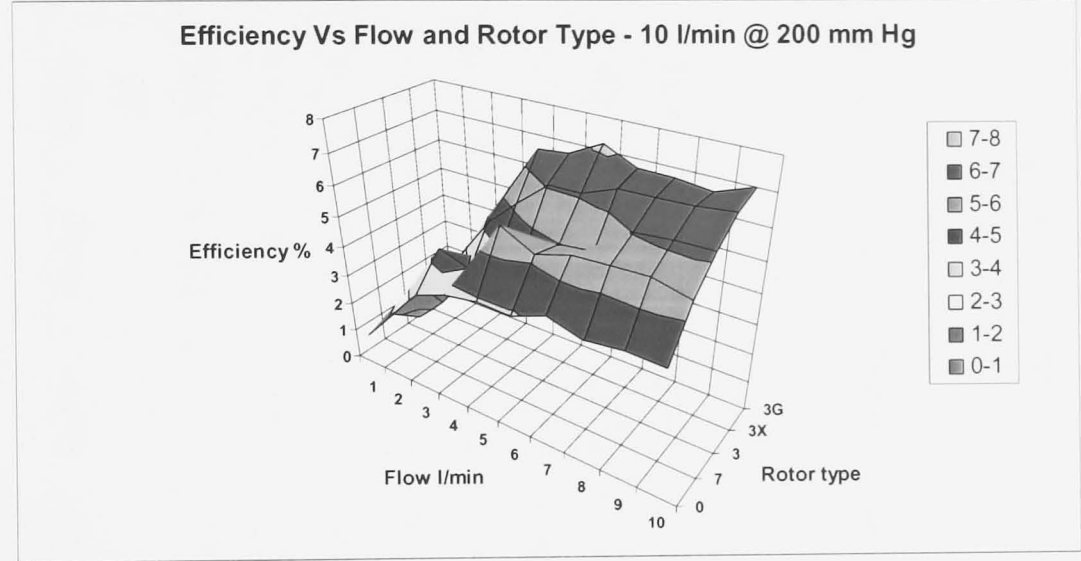


Figure 5.24. Three dimensional representation of efficiency Vs flow rate and rotor type at the setting point of 10 l/min @ 200 mm Hg.

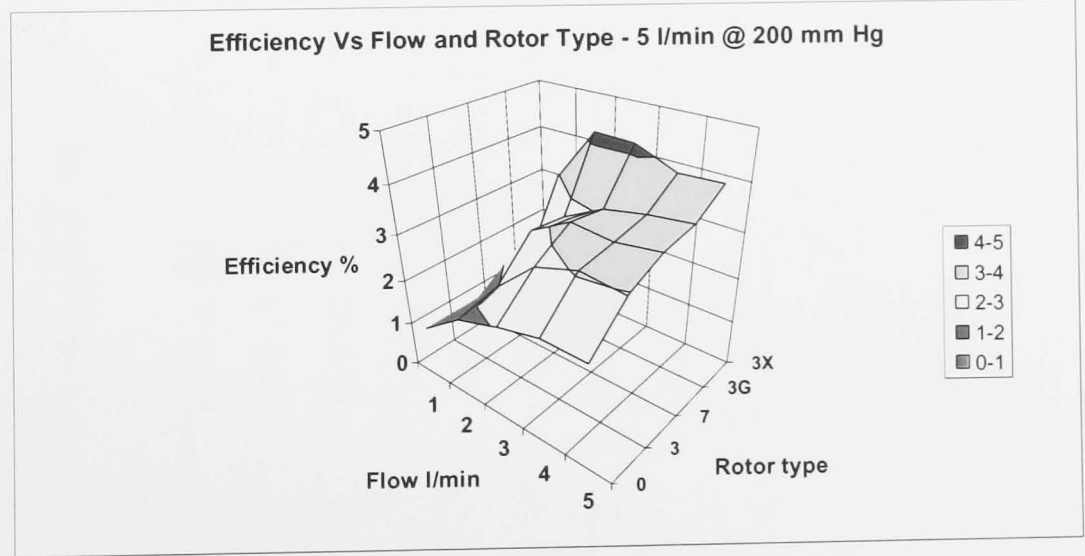


Figure 5.25. Three dimensional representation of efficiency Vs flow rate and rotor type at the setting point of 5 l/min @ 200 mm Hg.

Results of Flow Vs Volume Flow Rate Parameter, U_o

The results shown in Table 5.9 are shown for the 'set point' only.

Comparison no.	Performance 'set point'.	Rotor 0	Rotor 7	Rotor 3	Rotor 3G	Rotor 3B	Rotor 3X
1a	10 l/min @ 200 mm Hg	0.012	0.0774	0.0271	0.0295	0.0295	0.0272
2a	10 l/min @ 150 mm Hg	0.014	0.0942	0.0317	0.0346	0.0346	0.0323
3a	10 l/min @ 100 mm Hg	0.0166	0.1109	0.0389	0.0438	0.0438	0.0402
4a	5 l/min @ 200 mm Hg	0.0059	0.0402	0.0136	0.0135	0.0141	0.0139
5a	5 l/min @ 150 mm Hg	0.0069	0.0471	0.0155	0.0176	0.0176	0.0163
6a	5 l/min @ 100 mm Hg	0.0085	0.0568	0.0198	0.0218	0.0218	0.0203
7a	10 l/min @ both valves open.	0.0377	0.2611	0.0935	0.0987	0.0987	-
8a	10 l/min @ 200 mm Hg, inlet valve set / outlet valve wide open.	0.0133	0.0789	0.0272	0.0298	0.0298	0.0276
9a	5 l/min @ 100 mm Hg, inlet valve set / outlet valve wide open.	0.0085	0.0589	0.0205	0.0220	0.0220	0.0203

Table 5.9, The comparative results for Flow Vs Flow Rate Parameter U_o with the three rotors at the quoted 'set points'.

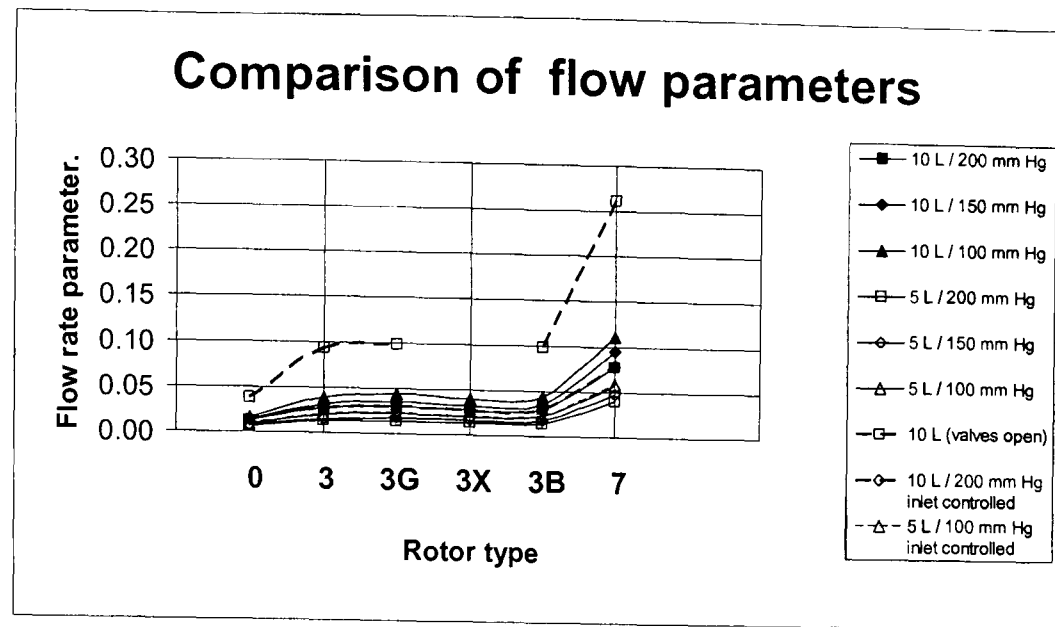


Figure 5.26 Showing the relationship between rotor type and the flow parameter at the 'set points' for all the fluids used.

Discussion of results, Flow parameter.

The flow parameter is assumed to be a dimensionless measure of the flow through the pump and that the higher the parameter the better the pump is running for a given performance.

From Figure 5.26, the rotor 7 had the highest flow parameter and that rotor 0 had the lowest flow parameter of the three rotors. This was taken as evidence that rotor build effects pump performance.

It was difficult to determine any differences between rotor 3, 3G, 3X and 3B indicating that viscosity has only a small effect on the flow parameter.

Examination of the results at 10 l/min @ 200 mm Hg, 10 l/min @ 150 mm Hg and 10 l/min @ 100 mm Hg indicate a slightly higher flow parameter with a reduction in pressure setting. This is confirmed by the result at 10 l/min with both valves open, where a high flow parameter is found irrespective of the rotor build compared to the other tests. This indicates that the lower the pump work rate, the higher the flow parameter.

It is difficult to determine any differences between both results at 10 l/min @ 200 mm Hg and both results at 5 l/min @ 100 mm Hg. It is assumed that the flow parameter is not affected if the pump delivery is controlled by throttling either the inlet or the output.

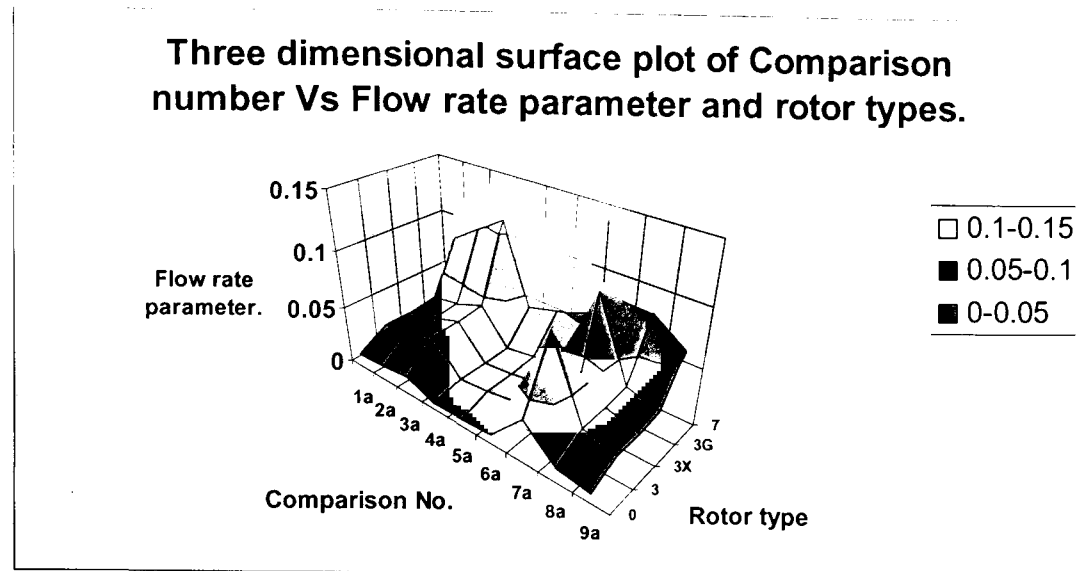


Figure 5.27. Three dimensional representation of Flow rate parameter Vs Flow and rotor type at the 'set points'

The values of the results at 0.02 to 0.05 agree with the calculated figures published in Crawford and Rice, (1974), although they are close to the low end of the range of 0.02 to 1.00 shown in the paper.

Figure 5.27 is a three dimensional representation of flow parameter Vs flow and rotor type at the 'set points' and indicates that the flow parameter varies only slightly with rotor build and performance setting. As the pressure setting reduces and the viscosity of the fluid increases, the flow rate parameter appears to increase. The peaks shown in Figure 5.27 are not understood

Results of Flow Vs Dimensionless Pressure Parameter P_t .

The results shown in Table 5.10 are for the 'set point' only.

Comparison no.	Performance 'set point'.	Rotor 0, Pressure parameter	Rotor 7, Pressure parameter	Rotor 3, Pressure parameter	Rotor 3G, Pressure parameter	Rotor 3B, Pressure parameter	Rotor 3X, Pressure parameter
1a	10 l/min @ 200 mm Hg	0.6820	0.8299	0.9019	0.9766	1.0183	0.912
2a	10 l/min @ 150 mm Hg	0.6780	0.9084	0.9265	1.0059	1.0488	0.961
3a	10 l/min @ 100 mm Hg	0.6795	0.8421	0.9269	1.0688	1.1144	0.993
4a	5 l/min @ 200 mm Hg	0.6785	0.8839	0.9131	0.9910	1.033	0.995
5a	5 l/min @ 150 mm Hg	0.7026	0.9081	0.8865	1.0429	1.0429	0.977
6a	5 l/min @ 100 mm Hg	0.7044	0.8780	0.9577	1.0657	2.791	1.015
7a	10 l/min @ both valves open	0.5935	0.8717	0.8653	0.9917	5.971	-
8a	10 l/min @ 200 mm Hg, inlet valve set / outlet valve wide open	0.6119	0.8548	0.9056	0.9925	3.126	0.941
9a	5 l/min @ 100 mm Hg, inlet valve set / outlet valve wide open.	0.7040	0.9414	1.0322	1.0882	3.580	1.011

Table 5.10, Comparative results for Flow Vs Stagnation Pressure parameter P_t with the three rotors and three fluids

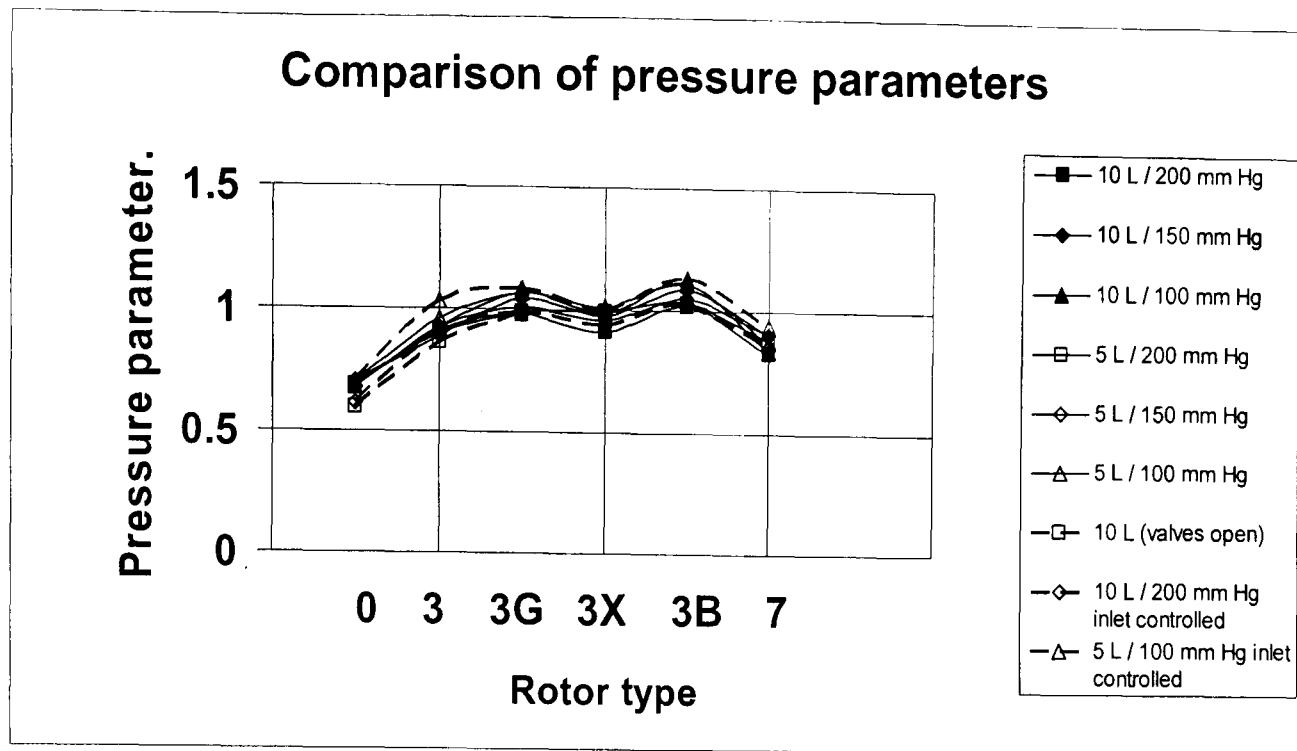


Figure 5.28. Showing the relationship between rotor type and pressure parameter at the set points for all the test fluids used.

Discussion of results, pressure parameter.

It is assumed that a pressure parameter is a dimensionless measurement of the pressure through the pump. From the results shown in Figure 5.28 above, it can be seen that the rotor 3 tended to have the highest pressure coefficient of rotors 0, 3 and 7 using water only. This confirms the concept of an 'optimum' rotor build for a particular fluid. The comparison of rotors 3 and 3G showed rotor 3G had the highest pressure coefficient indicating viscosity has an effect. The results for rotors 3 and 3X are similar which is not surprising considering the fluids tested had similar viscosity values.

Examination of the figures shows as the pressure setting for each test is lowered, i.e. the pump work rate reduces, the pressure parameter increases. The effect is small compared to the effect of rotor build and fluid viscosity.

It is difficult to determine the differences between both results at 10 l/min @ 200 mm Hg and both results at 5 l/min @ 100 mm Hg.

The values of the pressure parameter obtained from the tests, 3.58 to 0.912, are similar to the calculated figures quoted in Crawford and Rice,

(1974), which range from plus 3.5 to negative values. It is not understood how a pump can have a negative pressure parameter.

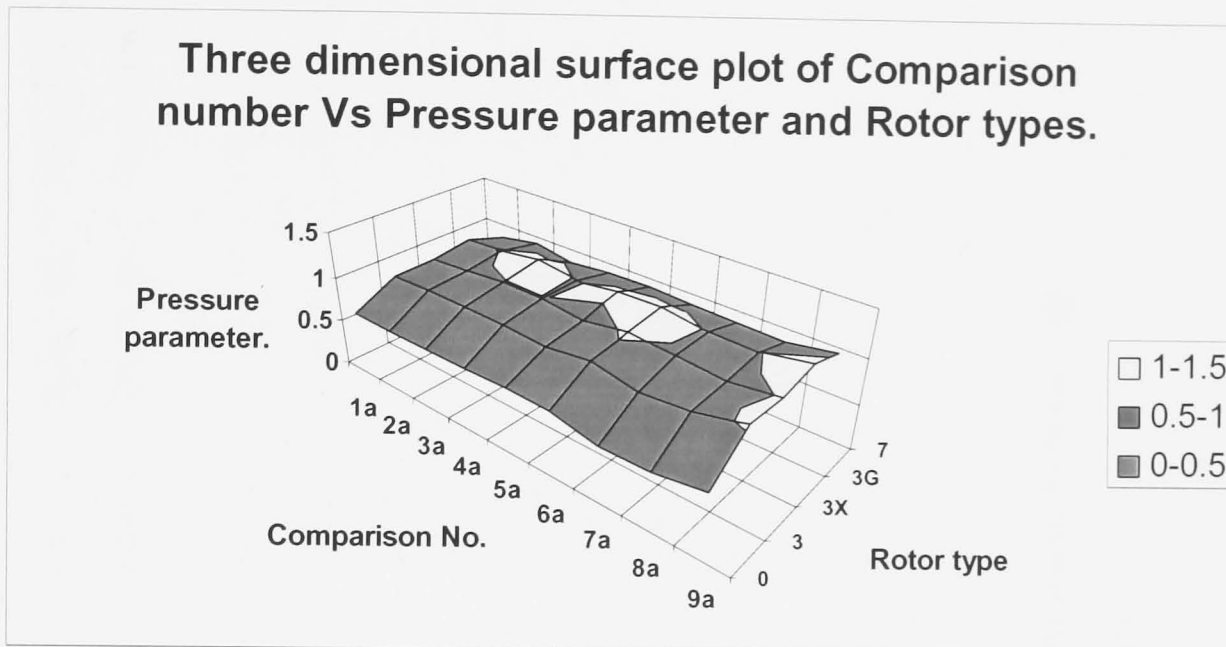


Figure 5.29. Three dimensional representation of Pressure parameter Vs Flow and rotor type at the 'set points'

Figure 5.29 is a three dimensional representation of the pressure parameter Vs flow and rotor type at the 'set points'. It shows that the pressure parameter varies only slightly with rotor build and performance setting, rotor 0 having the worst result of all the rotor and fluid combinations tested.

Results of Flow Vs Reynolds number.

The figures shown here are a representation of the Reynolds number and are used for comparative purposes only. The calculation is taken from the paper by Crawford and Rice, (1974), and is understood to represent the Reynolds number of the fluid passing between a pair of adjacent discs and not through the pump itself. The 'best' performance was considered to be lowest Reynolds number achieved.

The results shown in Table 5.11 are for the 'set points' only. For the purposes of plotting, the values for rotor 0 are reduced by a factor of 10^2 , otherwise it would not be possible to see the results for the other values on the same graph.

Comparison no.	Performance 'set point'	Rotor 0, Reynolds no.	Rotor 7, Reynolds no.	Rotor 3, Reynolds no.	Rotor 3G, Reynolds no.	Rotor 3B, Reynolds no.	Rotor 3X, Reynolds no.
1a	10 l/min @ 200 mm Hg	240997	95	3256	624	903	3238
2a	10 l/min @ 150 mm hg	208546	78	2783	532	769	2732
3a	10 l/min @ 100 mm Hg	170532	66	2270	421	609	2194
4a	5 l/min @ 200 mm Hg	240997	91	3241	619	895	3167
5a	5 l/min @ 150 mm Hg	203819	78	2841	522	755	2413
6a	5 l/min @ 100 mm Hg	166908	65	2232	422	610	2169
7a	10 l/min @ both valves open.	74916	28	944	187	270	-
8a	10 l/min @ 200 mm Hg, inlet valve set / outlet valve wide open	212731	93	3250	619	895	3194
9a	5 l/min @ 100 mm Hg, inlet valve set / outlet valve wide open	166641	62	2153	419	606	2171

Table 5.11, The comparative results for Flow Vs Reynolds Number with the three rotors and three fluids.

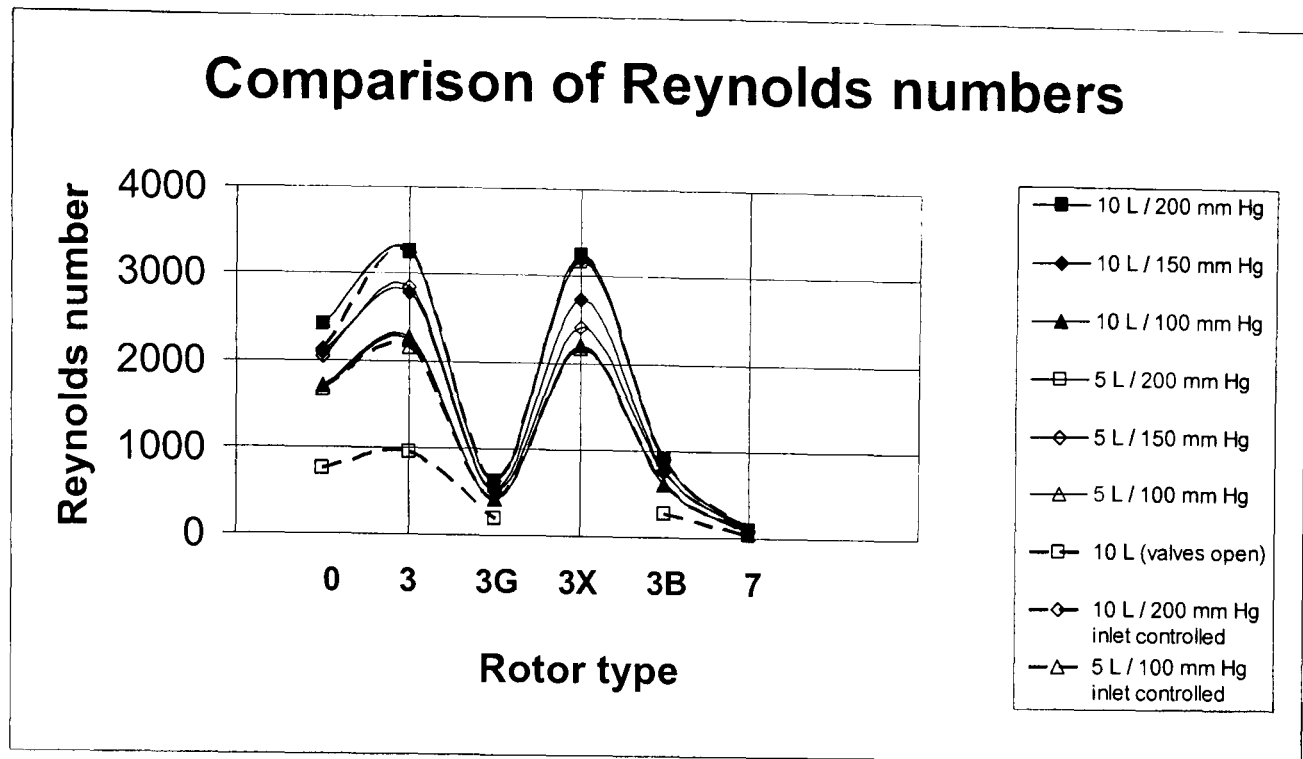


Figure 5.30. Showing the relationship between rotor type and Reynolds number at the 'set points' for all the test fluids used.

(Note: - Rotor 0 values are reduced by a factor of 10^2)

Discussion of results, Reynolds number.

The value of the Reynolds number can be a critical factor in determining the success or failure of any fluid flow application. Massey, (1989), Chapter 5, 'Two Kinds of Flow', indicates there is a 'limiting' Reynolds number that determines whether fluid flow is laminar or turbulent.

For water flow through pipes under ordinary conditions, the limiting Reynolds number between laminar and turbulent flow is usually taken as 2000. This number is not precise in the sense that flow at a Reynolds number of 1999 is laminar and changes abruptly to turbulent flow at a Reynolds number of 2001 but the value of 2000 should be used as a guide to the flow conditions of a fluid only for flow in a pipe.

For the purposes of a blood pump, it is considered that the lower the Reynolds number, the more the fluid is moving towards laminar flow, where laminar flow is assumed to have the least fluid stresses and less potential for blood damage. For the purposes of these tests the 'best' performance is considered to be the result with the lowest Reynolds number irrespective of the value of that number.

If all the figures from Table 5.11 above were plotted on the same scale on Figure 5.30, the results for rotors 3 to 7 could not be differentiated from one another due to the high values for rotor 0. Accordingly the figures for rotor 0 are plotted on a scale reduced by a factor of 10^2 compared to the other results and are included for completeness only.

From Figure 5.30, rotor 7 had the lowest Reynolds number of the three rotors using water or any of the other test fluids. This result indicates that rotor build is important with this pump type and may be connected with the width of the gaps between the discs. It indicates that there could be an optimum rotor build for a particular application.

From Figure 5.30 the comparison of rotors 3 and 3G show that rotor 3G had the lowest Reynolds number of the two. This result shows that the Tesla pump is sensitive to fluid viscosity and that the more viscous the fluid, the better the pump operates in terms of Reynolds numbers

Rotors 3B and 3G have similar values. Rotor 3B is a calculated result using a figure for the viscosity for blood which is close to the viscosity for the water / glycerine mixture.

Rotor 3 and rotor 3X have similar values of Reynolds numbers, as expected, as the viscosities of the two fluids are similar.

Assuming that the Reynolds number indicating the change from laminar flow to turbulent flow is 2000, rotors 3G, 3B and 7 would appear to be in laminar flow as they are below 2000. Rotors 0, 3 and 3X would appear to be turbulent flow as they are above 2000.

Examination of Table 5.11 indicates that the results tend to form a pattern. Results for comparisons 1a, 4a and 8a have values of the Reynolds number that are very similar. The values of the Reynolds numbers for comparisons 2a and 5a are close together as are the values for comparisons 3a, 6a and 9a. The results for comparisons 1a and 8a and for comparisons 6a and 9a appear to be almost identical with each other. The relationship in each case appears to be the pressure setting at the 'set point'.

These figures cannot be compared with the Reynolds numbers obtained from the paper by Crawford and Rice, (1974), as the maximum figure quoted in the paper is 10.

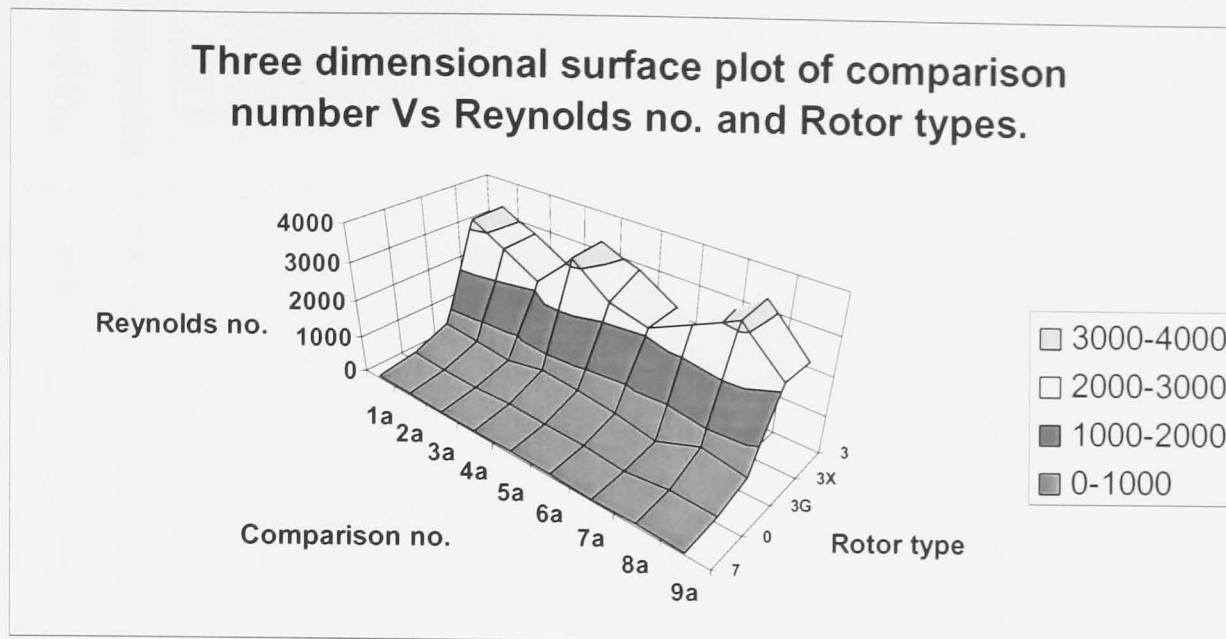


Figure 5.31. Three dimensional representation of Reynolds number Vs Flow and rotor type at the 'set points'.

Figure 5.31 is a three dimensional representation of the Reynolds numbers Vs flow and rotor type at the 'set points'. It shows that the Reynolds number varies with rotor build and performance setting. The result for rotor 0 should be ignored as it is not comparable on the scale shown. Figure 5.31 confirms that rotor 7 has a low Reynolds number irrespective of the comparison number. Comparing the results of rotor 3 in all its applications, the fluid with the highest viscosity, rotor 3G, shows the lowest Reynolds number while, rotors 3 and 3X, have the highest Reynolds number and are very similar to each other. The viscosity difference between the 3 and 3X was small and this result is not unexpected. The effect of workload on the pump is reflected in the difference in Reynolds numbers between the various comparison numbers where the 'peaks' of the plot indicate the highest pressure settings and the 'troughs' indicate the lowest pressure settings.

Results of Flow Vs Dimensionless Torque Parameter T_1 .

The results shown in Table 5.12 are shown for the 'set point' only.

Comparison no.	Performance 'set point'.	Rotor 0, Torque parameter	Rotor 7, Torque parameter	Rotor 3, Torque parameter	Rotor 3G, Torque parameter	Rotor 3B, Torque Parameter	Rotor 3X, Torque parameter
1a	10 l/min @ 200 mm Hg	1.220	8.081	2.574	2.654	2.767	2.307
2a	10 l/min @ 150 mm Hg	1.201	7.083	2.608	2.702	2.817	2.33
3a	10 l/min @ 100 mm Hg	1.208	8.073	2.625	2.793	2.913	2.389
4a	5 l/min @ 200 mm Hg	1.220	6.868	2.730	2.842	2.963	2.153
5a	5 l/min @ 150 mm Hg	1.082	6.675	2.609	2.911	3.035	2.185
6a	5 l/min @ 100 mm Hg	1.135	7.392	2.773	2.677	2.791	2.171
7a	10 l/min @ both valves open	2.197	17.462	5.449	5.727	5.971	-
8a	10 l/ min @ 200mm Hg, inlet valve set / outlet valve wide open	1.900	6.9598	2.653	2.998	3.126	2.199
9a	5 l/min @ 100 mm Hg, inlet valve set / outlet valve open.	1.155	7.991	2.741	3.434	3.58	2.275

Table 5.12, The comparative results for Flow Vs Torque Parameter T_1 with the three rotors at the quoted 'set points'.

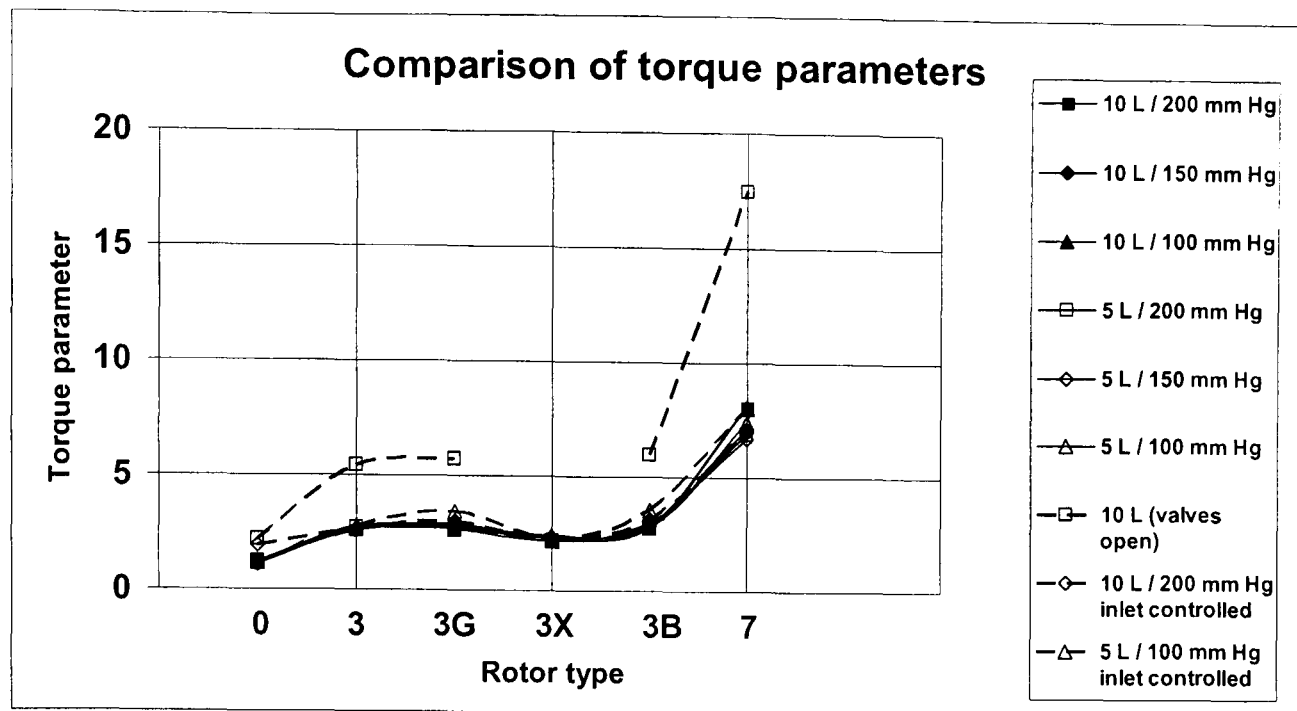


Figure 5.32. Showing the relationship between rotor type and the torque parameter at the 'set points' for all the test fluids used.

Discussion of results, torque parameter.

It is assumed that the torque parameter is a dimensionless measure of the power required to operate the pump and that the higher the torque parameter the better the pump is operating for a given performance.

Figure 5.32 compares rotors 0, 3 and 7, and shows that rotor 7 had the highest torque parameter and rotor 0 had the lowest parameter of the three rotors using water only. This result indicates that the rotor build affects the torque parameter. A comparison of rotor 3 with rotor 3G showed that rotor 3G had a torque parameter that is marginally greater than rotor 3 indicating that viscosity has only a small effect. The results for rotors 3G and 3B and rotors 3 and 3X are almost identical to each other as expected, as the viscosities in each case are almost identical.

The results of comparisons of 10 l/min @ 200 mm Hg and 10 l/min @ 200 mm Hg (inlet controlled) and 5 l/min @ 100 mm Hg and 5 l/min @ 100 mm Hg (inlet controlled), show that they appear to be almost identical. This indicates that there is little or no change in the torque parameter if the pump output is controlled by throttling the inlet or outlet.

The values of the torque parameter obtained from the tests, 17.46 to 1.08, are within the range of 0 to > 1000 in the paper published by Crawford and Rice, (1974).

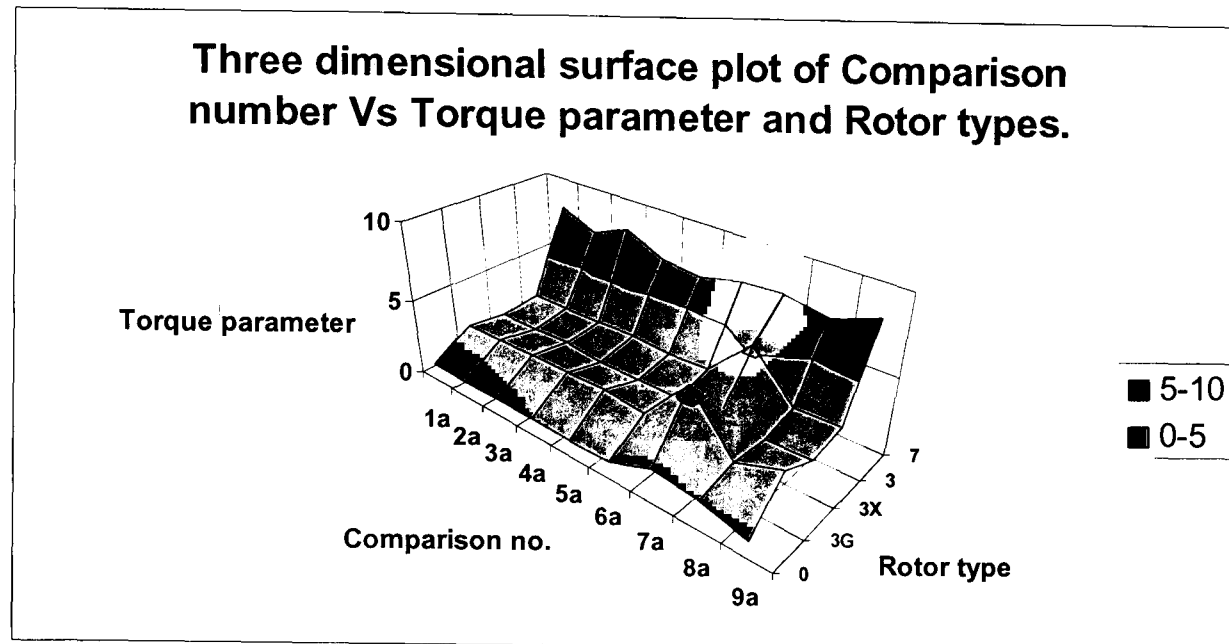


Figure 5.33 Three dimensional representation of Torque parameter Vs Flow and rotor type at the 'set points'

Figure 5.33 is a three dimensional representation of the calculated torque parameter Vs flow and rotor type at the 'set points'. It indicates that the torque parameter varies noticeably with rotor build, as seen by the difference in parameter with rotors 0 and 7, but changes very little with performance setting. There does not appear to be much effect on torque parameter with the change in fluid viscosity.

5.7, Flow visualisation.

5.7.1 Introduction.

A video was made to show the fluid flow pattern inside the pump from various angles and at various flow rates. How the fluid passes through the pump is important in terms of identifying points where high shear stresses could be generated and if any blood recirculation or stagnant flow areas are found inside the pump, stress and surface reactions may set up thrombi which could then pass through the pump causing problems to the patient.

A separate part of the video showed general views of the rig and how it operated.

Two fluids are shown in the video. The first fluid was plain water and the second was the water / glycerine mixture described previously.

All the sequences were taken with the pump fitted with rotor 3.

Observations were made from three directions in order to determine the flow patterns. These directions were: -

- a) Along the axis of rotation from the fluid inlet side of the pump,
- b) From above, looking down on the discs and
- c) Along the outlet passage in the direction of flow with the fluid moving away from the camera.

Figure 5.34 shows this arrangement.

In order to see the flow patterns, air was injected into the pump inlet and a light source was used to illuminate the pump such that the air bubbles would show up against a black background placed behind the pump.

Still photographs, (35mm), were taken as necessary during these tests and are shown in Figure 5.40 onwards.

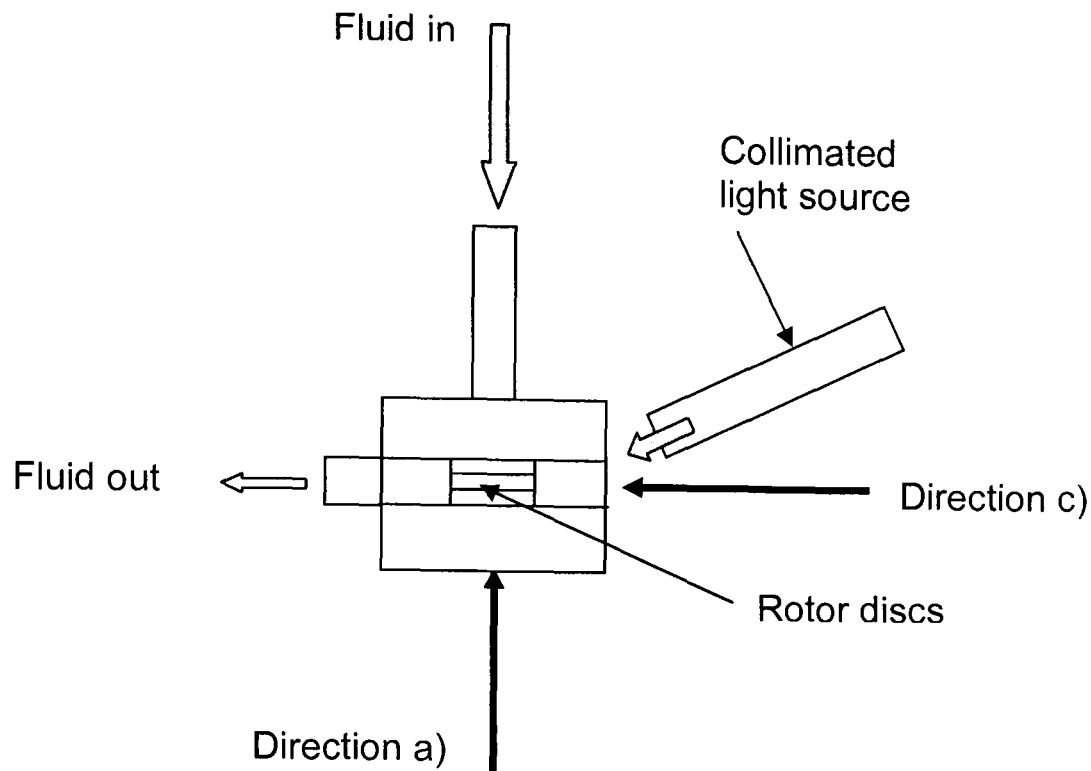


Figure 5.34 Layout of directions used to view the fluid flow.

Direction a) is looking along the axis of rotation,

Direction b) is looking down on the discs (as shown above).

Direction c) is looking along the outlet passage.

5.7.2 Video with water only.

The pump was set up to deliver 10 l/min at a suitable delivery pressure. It was ensured that no air bubbles were seen in the water flow inside the pump and the light source was adjusted to illuminate the water inside the pump volute.

Observations from direction a).

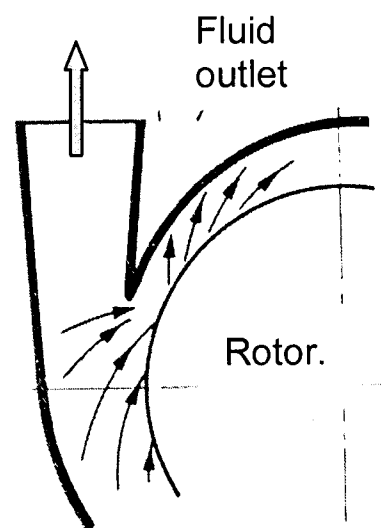
With the pump running at the set condition and with no air present in the fluid, air was deliberately injected into the fluid inlet in order to view the characteristics of the fluid flow. The air initially appeared as a white 'fog' inside the pump but as the pump ran the 'fog' began to clear when it was noticed that a vortex, comprising air bubbles, had been generated in the water flow. The vortex was near the cutwater and covered the vertical distance from the

periphery of the discs to the upper wall of the pump outlet. As time went on, the vortex moved back and forth across the discs and eventually decayed as the amount of air bubbles reduced.

The reason for the vortex is not known but it was assumed that it was an effect of the water being picked up and rotating with the disc(s) against the stationary wall of the volute at the entry position.

It may be possible to reduce this effect by improving the fluid entry conditions and suggestions are shown later in this report.

It was noted, following the movements of the air bubbles, that part of the flow in the volute tended to 'dive' under the cutwater and part to be swept along the flow passage where it dispersed. This was very pronounced at times and the flow after the cutwater along the outlet passage tended to have a recirculating flow pattern. This was more noticeable at low flow rates and pressures. Lazarkiewicz and Troškolański, (1965), indicate that these effects are not unknown and are connected with a non uniform pressure distribution around the inner wall of the volute casing which in turn causes eddying and back flow around the area of the cutwater. This effect is most pronounced when the pump delivery approaches zero. Figure 5.35 below illustrates the effect from Lazarkiewicz and Troškolański, (1965), and Figure 5.36 is a representation of the actual effect seen during the pump performance.



Lazarkiewicz S and Troškolański, A, (1965)

Figure 5.35, The flow pattern at the cutwater
at low delivery.

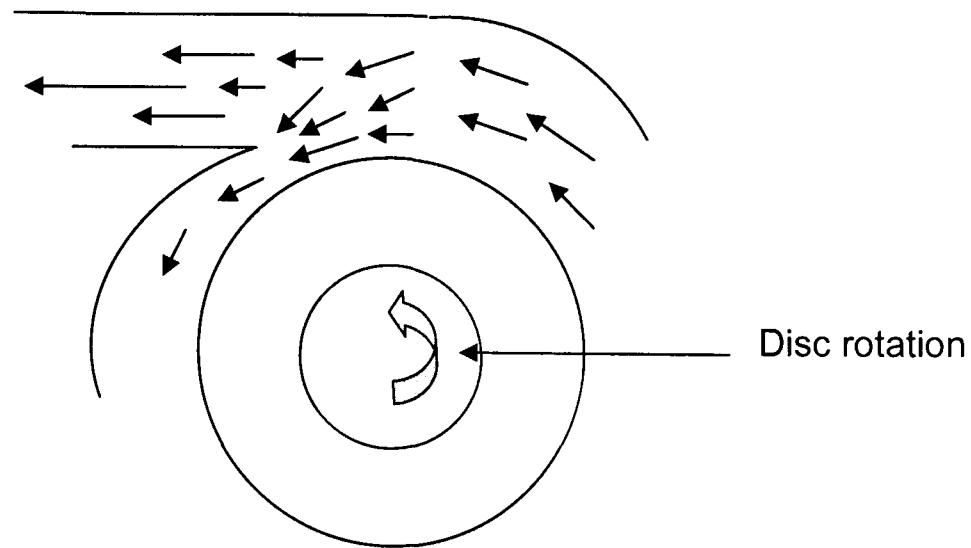


Figure 5.36. Flow 'diving' under the cutwater, viewed from direction a), axis of rotation.

Observations from directions b and c).

The vortex tended to form at the cutwater and was mainly up against the volute wall closest to the inlet but occasionally jumped across the rotor to the gap between the first two discs, a distance of about 7 mm. The height of the vortex at this point was about 18 mm, the distance from the edge of the discs to the top wall of the outlet volute. Figures 5.37, 5.38 and 5.39 are a representation of what was happening.

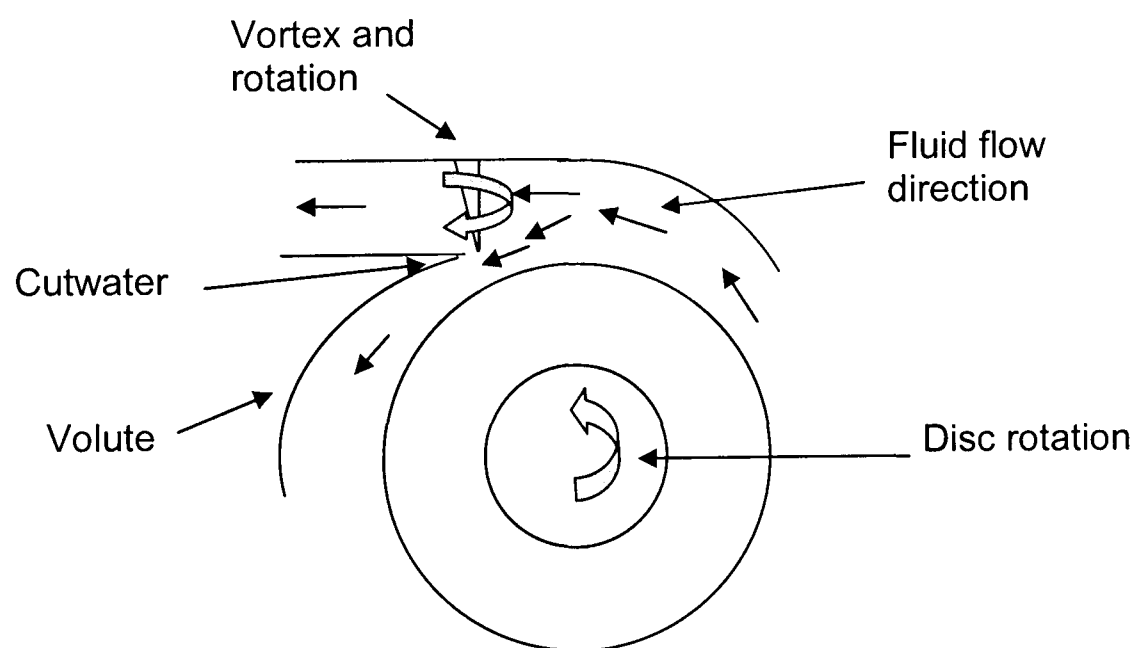


Figure 5.37. Vortex position from direction a), axis of rotation.

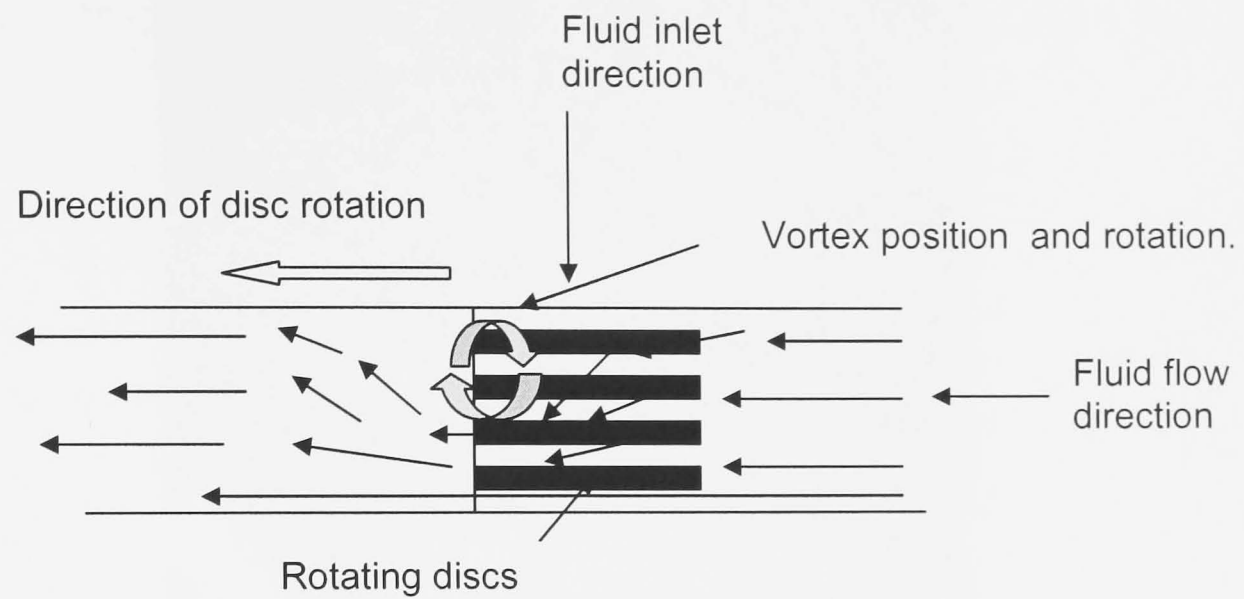


Figure 5.38, Vortex position from direction b).

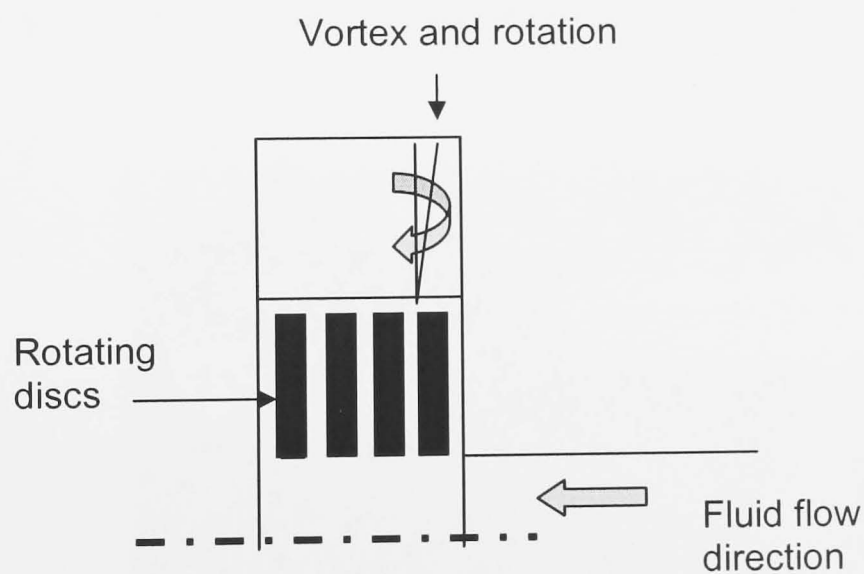
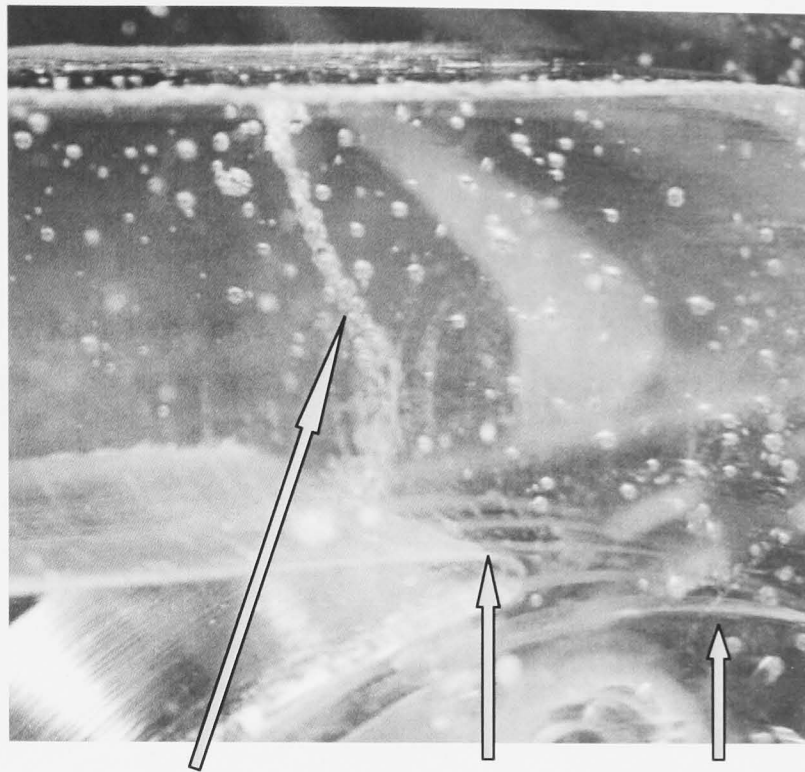


Figure 5.39. Vortex position from direction c).

Two photographs of the vortex were taken, Figures 5.40 and 5.41. These show the column of bubbles indicating the position of the vortex that appeared close to the cutwater and to the inlet side of the pump.



Column of air bubbles. Cutwater. Rotor.

Figure 5.40. Column of air bubbles showing the vortex at the cutwater.

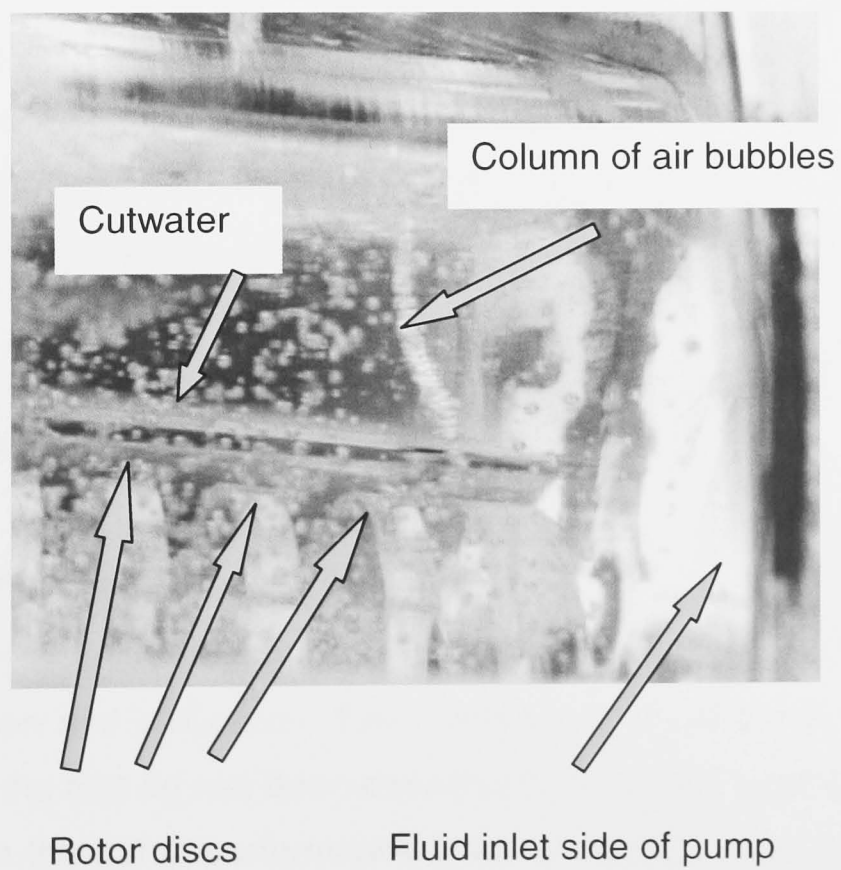


Figure 5.41 Column of bubbles showing the vortex relative to the discs.

Further work is needed to improve the hydraulic performance of the pump. The application of a Computational Fluid Dynamics, (CFD), analysis package should provide a solution.

5.7.3 Video with the water and glycerine mixture.

A second set of video images was taken with the water and glycerine mixture. The pump was set up to give a specific performance for each video sequence.

A similar performance to the water only tests was noted regarding the generation of the vortex and its position in relation to the cutwater and flow passage of the pump. The vortex occurred in the same place at the cutwater and across the pump flow passage as for the water only tests.

It was noted that the flow pattern immediately after the cutwater was not stable. The flow pattern rotated clockwise at one time and anti-clockwise at another.

With the pump stationary it was noticeable that the air bubbles trapped in between the discs did not disperse as quickly with the water / glycerine mixture compared to water only. One sequence, taken after the tests at the prepared conditions had been run, showed this clearly with the bubbles rising slowly from between the discs to the top of the flow passage in the pump. This is not unexpected as the water - glycerine mixture has a higher viscosity than water.

The rest of the video shows general views of the test rig and of the instrumentation and how they were operated.

5.7.4 Summary

The construction of the test rig has been outlined with details of the instrumentation and calibration. The construction of the pump, rotors and the assembly to the test rig has been described and a brief description of the test fluids used in the pump performance tests is included. The water - glycerine mixture reproduced a fluid that gave a viscosity equivalent to blood but at only one temperature. It was not possible to produce a test fluid that accurately modelled blood viscosity over a wide temperature range. The water – xanthan

gum fluid was not useful for simulating blood. Although mixed to the strengths quoted by Wickramasinghe, (2002), the performance of this fluid was not appreciably different to that of water. There does not appear to be a fluid that can accurately model blood characteristics and it is not understood why an industry involved in the production of blood contacting medical devices does not have a standard, clearly specified, analogue fluid that can accurately model blood for use with the development of its products. This implies that different manufacturers may not be working to the same standard and that results may not be comparable.

The results of the tests showed that the Tesla pump was capable of pumping the target amounts of fluid at the target pressures. There was a difference between the performance of the rotors, where the rotor with the smallest number of discs, rotor 0, gave the worst performance. The results from all the rotors indicated that there may be an optimum rotor build dependant on fluid viscosity but this needs more investigation. The results of the constant speed tests showed that the Tesla pump maintained the pressure, as set, over the delivery range of 1 – 10 l/min. and that the pressure appeared to be dependant on rotor speed. If rotor speed is translated into peripheral velocity, a small Tesla pump, running at high speed, may be able to produce the same pressure as a large Tesla pump running at a lower speed. Further work is necessary to confirm this point.

Estimates of blood stress in the gap between the periphery of the disc closest to the volute wall and the wall itself showed a maximum stress of 29 N/m. Calculations indicated that this figure could be lower with design changes to the pump.

Power consumption ranged from 120 to 4 Watts depending on the speed of the pump. This was the power required to pump the fluid and did not include the power to overcome friction.

The pump efficiency was low, the maximum obtained was 10–12%. The reason for this is not understood but small pumps are known to be less efficient as friction and fluid losses are greater in proportion compared to large pumps.

The non-dimensional parameters were comparable to previous research by Crawford and Rice, (1974), although tended to be at the one end of the ranges quoted.

As found from practical experience, the Tesla pump Reynolds numbers looked reasonable but were higher than the theoretical results quoted in the paper by Crawford and Rice, (1974). Using the formula in the paper and using the speeds used with the Tesla pump tests, the gap between the discs was calculated at 0.05 to 0.3 mm. This is smaller than the gaps used in these tests and may be the reason why the Reynolds number results are so different.

A problem was found with the fluid flow pattern inside the pump where a vortex was found inside the fluid close to the cutwater. The problem may be caused by the closeness of the end disc to the wall of the volute at fluid entry and it may be possible to reduce or eliminate the problem by changes to the pump build. Increasing the gap between the end disc and the volute wall may change the output of the pump but would reduce the fluid stress levels at that point.

Work is necessary to eliminate this problem. The application of a Computational Fluid Dynamics, (CFD), package should be able to provide a solution.

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American Chemical Society and American Institute of Chemical Engineers.

Chapter 6, Discussion.

6.1 Tesla pump performance.

As there does not appear to be a published specification for blood pumps to assist or replace the human heart, and published data rarely includes engineering specifications and performance curves, it is difficult to compare the output of the pumps produced to date with what is required. The best that can be done is to compare the results already published and assume the performances found are suitable for medical purposes.

The physical characteristics of a sample of blood pumps are shown in Table 2.10. This compares the volume pumped, the installed volume and the mass of each pump together with the pump type as far as can be determined from the available literature. These are compared to the human heart which has been given a value of 1 for each of the characteristics.

Table 2.10 indicates that blood pumps can be placed in three groups. The Total Artificial Hearts and pulsating pumps are the largest and heaviest and the axial flow pumps are the smallest and lightest. The rotary pumps fall in between these two groups in terms of size and weight. There also appears to be a difference in delivery between the pumps that are already in use, (the commercial pumps), and what appear to be experimental pumps. The commercially produced pumps deliver similar amounts of blood but the experimental pumps deliver amounts that appear to be either too small or too large for clinical use. It is assumed that these pumps have not, or will not, reach hospital use and are produced only as investigative models.

From the published papers, it is difficult to determine the performance of the published pumps described in this thesis, as only single point flow Vs pressure figures are given, for example, 10 l/min at a pressure of 100 mm Hg. It is assumed that the pressure quoted is the delivery pressure.

The performances from Table 2.10 have been recalculated and compared with the Tesla pump by plotting the single point flow Vs pressure figures for the published pumps against the setting points for the Tesla pump. Only the extremes of the performance of the Tesla pump as tested are plotted as any combination of flow, from 1 to 10 l/min and pressure, from 1 to 200 mm

Hg, can be covered by the Tesla pump as shown by the results in this thesis.

For the purposes of this plot the human heart delivery is taken as 5.5 l/min at a pressure of 110 mm Hg. which is midway between the quoted systolic and diastolic pressures of 140 / 80 mm Hg. These results shown in Figure 6.1, indicate that some pumps can give a performance superior to the human heart. This is seen for some of the pulsating and for some of the rotary radial pumps. The rotary axial pumps appear to be equivalent to or slightly less than the heart. The Tesla pump can give a performance equivalent to, or greater than, the maximum flow requirements of the published pumps at 5 to 10 l/min at pressures of up to 200 mm Hg.

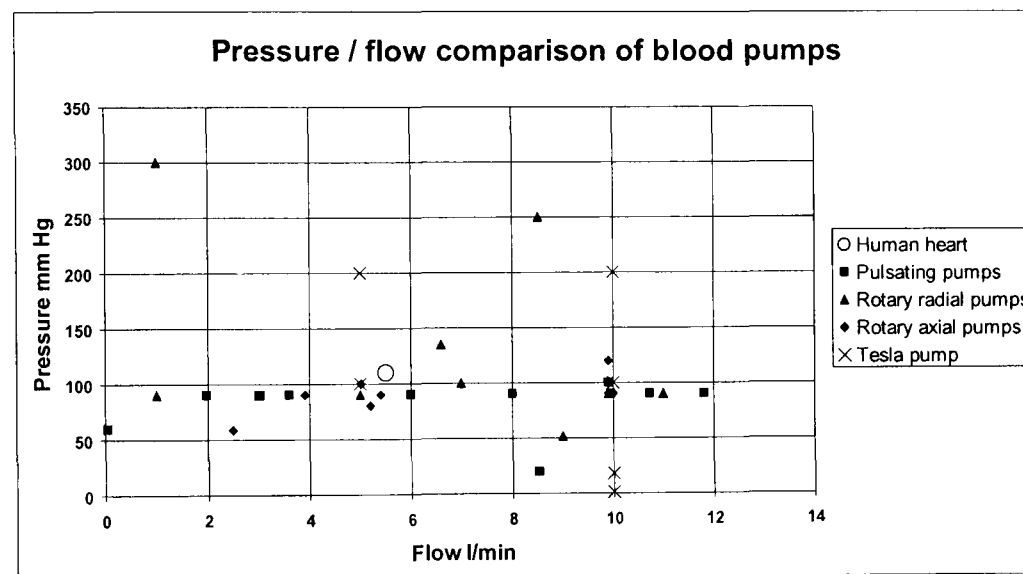


Figure 6.1. Comparison of published pump results with the Tesla pump.

One result for the Tesla pump is shown where both inlet and outlet valves were set fully open. Two pressures are shown, the delivery pressure at 1 mm Hg, the differential pressure at 18 mm Hg both at a delivery of 10 l/min.

These results indicate that the Tesla pump has a wide performance envelope with deliveries up to 10 l/min. at pressures ranging from 1 to 200 mm Hg. There is no information to show that any of the other pumps can do this and this would make the Tesla pump more versatile than some of the other types in terms of performance requirements.

The Tesla pump may be capable of exceeding the performance shown. It did not appear to be running at the maximum possible output under the above conditions as an amount of adjustment was available at the valves

controlling the flow and the speed control system was not at the maximum setting.

To provide a clearer comparison, the above result was replotted as the ratio's of the performance of the human heart and the blood pumps and is shown on Figure 6.2

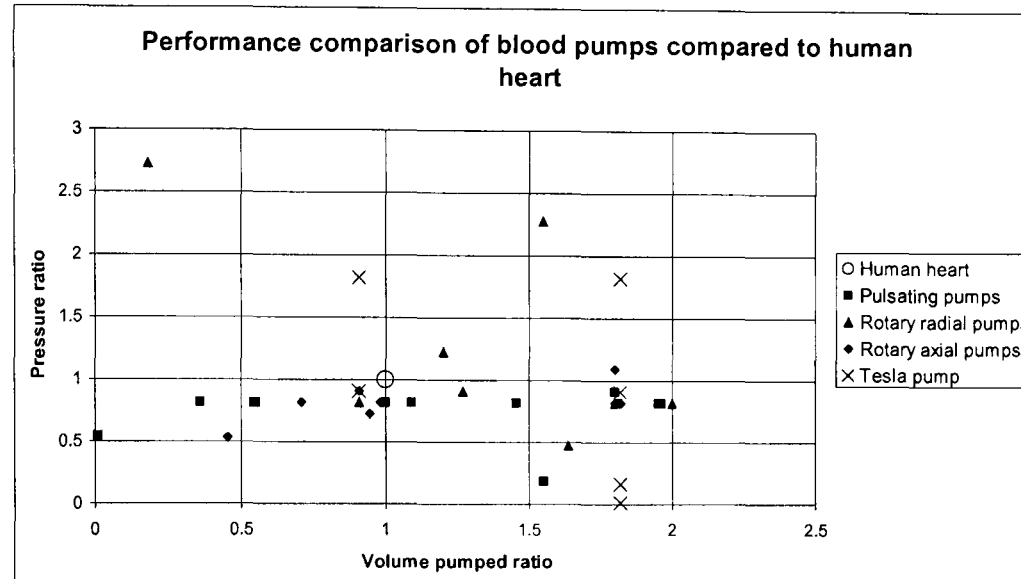


Figure 6.2. Comparison of published volume and pressure results with the Tesla pump shown as a ratio compared to the human heart.

This figure confirms that pulsating pumps are capable of flows approaching twice that of the human heart but do not appear to provide pressures much different to the heart. Rotary radial pumps can provide approaching 2½ times the pressure and over 1½ times the delivery of the heart. Rotary axial pumps can provide up to twice the delivery but can only equate the pressure of the heart.

The Tesla results envelope indicates that the pump is capable of up to twice the delivery and up to twice the pressure of the heart which must be a clear advantage compared to the other pumps.

An examination of the mass of each blood pump compared to the human heart is shown on Figure 6.3. This shows that the pulsating pumps can be up to three times heavier than the heart. Some of the rotary radial pumps are heavier than the heart, one of them up to three times, but the rotary axial pumps are equivalent to, or, lighter than the heart.

The Tesla pump was weighed. It is approximately four times heavier than the heart. This is not considered to be important at this time, as the pump

has had no work conducted on it to make it suitable for medical applications. With a proper design and use of materials it should be able to reduce this to at least 50% of the present pump. This would bring the mass of the Tesla pump into the region of the pulsating and rotary radial pumps but it would be unlikely to make it comparable to the mass of the rotary axial pumps.

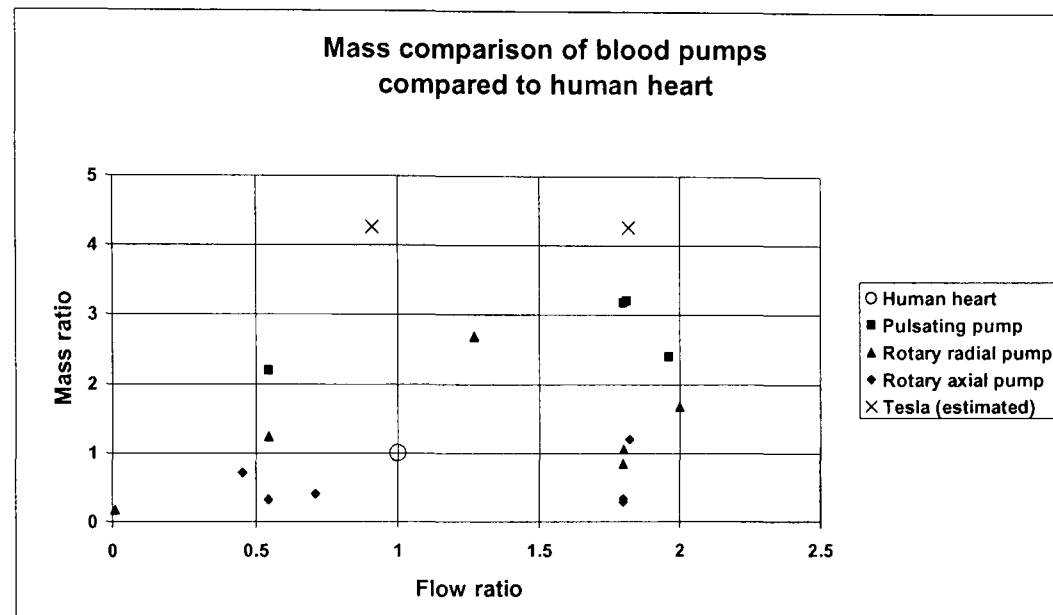


Figure 6.3. Comparison of mass and flow ratio results with the Tesla pump shown as a ratio compared to the human heart.

The published pumps were also compared in terms of the installed volumes as shown on Figure 6.4. All the published pumps appear to have a smaller volume than the human heart. A similar pattern compared to the previous result was noticeable in that the pulsating pumps had the largest volumes followed by the rotary radial pumps with the rotary axial pumps appearing to be the smallest.

An estimation of the installed volume of the Tesla pump made it slightly greater than the human heart. As before this figure for the volume of the Tesla pump is not considered to be very important as there has been no work conducted on it to make it suitable for clinical application. It is expected that the volume could be reduced to at least the level of the rotary radial pumps which would make it about 25% of the volume of the human heart.

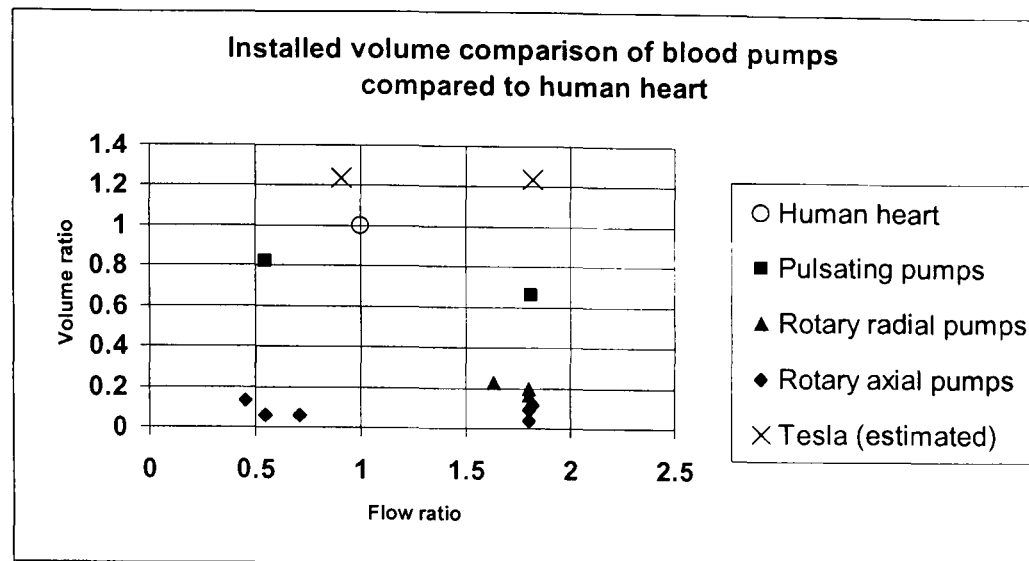
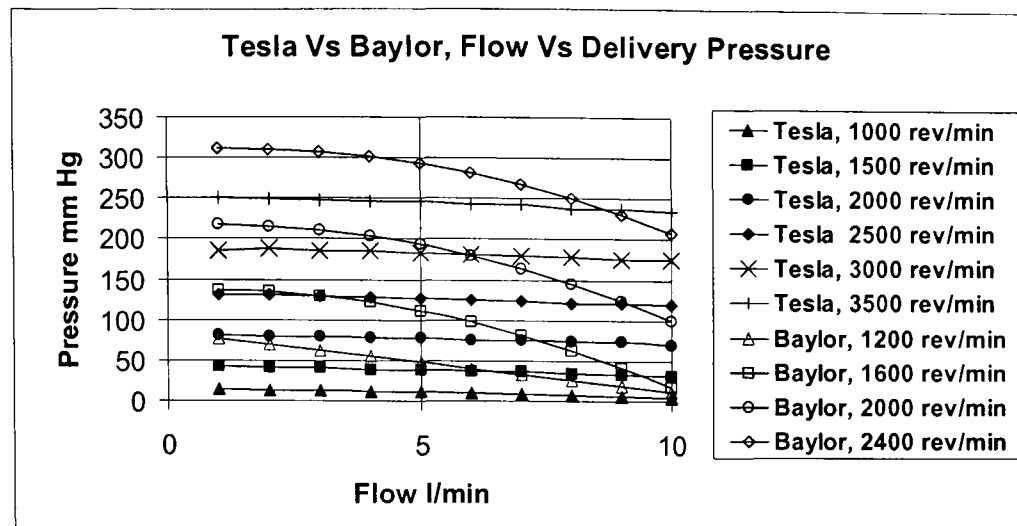


Figure 6.4. Comparison of published pump results with the Tesla pump shown as a ratio compared to the human heart

6.1.1 Constant speed performance.

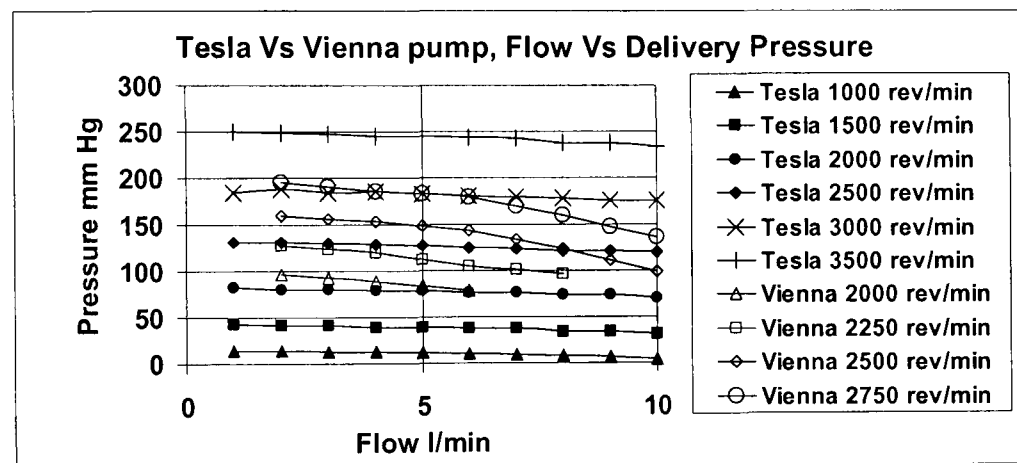
A literature search was made to determine the constant speed performance comparisons between the Tesla pump and other pumps. Three sets of constant speed results were found, two were for centrifugal, rotary pumps, the Baylor and the Vienna pumps, (similar designs to the Tesla pump), and the third result was for an axial flow pump, the Nimbus pump. The Tesla pump was tested by setting at the required speed and the output was adjusted to give a delivery of 10 l/min. The results are compared on Figures 6.5, 6.6 for the centrifugal pumps and Figure 6.7 for the axial flow pump. There were no comparative results found for a pulsating pump that covered a delivery change relative to a change in 'beat' rate.

The pressures shown here for the Tesla pump are delivery pressures and it is assumed that the pressures for the Baylor pump, the Vienna pump and the Nimbus pump are the same.



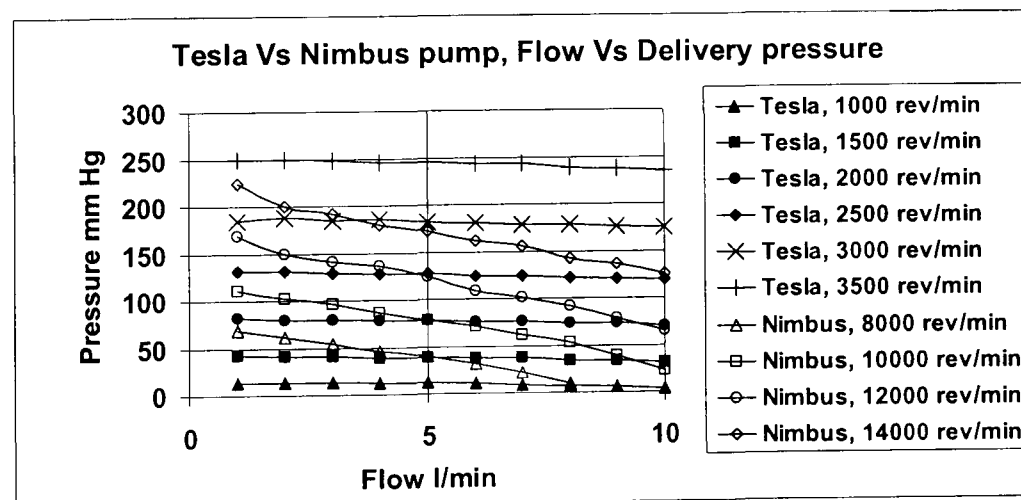
O'Hara, Makinouchi, Orime et. al. Artif. Organs, Vol. 18, No. 1, 17-24, 1994.

Figure 6.5. Comparison of Tesla and Baylor pump constant speed results.



Schima, Schmallegger, et. al., Art. Organs, 19(7); 639-643, 1995.

Figure 6.6. Comparison of Tesla and Vienna pump constant speed results



Butler, Dow, et. Al., (1999). Ann. Thorac. Surg., 1999; 68: 790-4.

Figure 6.7. Comparison of Tesla and Nimbus pump constant speed results.

These results show that the Tesla pump maintains a relatively constant pressure, over the flow range tested at constant speed, compared to the other pumps. Taking the Tesla pump result at 3500 rev/min, the pressure difference over the delivery range of 1 to 10 l/min is 4.5 mm Hg for the differential pressure and 15.4 mm Hg for the delivery pressure. With the other pumps, the pressure drop is greater than this as the flow increases. The estimated delivery pressure drops for a flow of 1 to 10 l/min are 100 mm Hg for the Baylor pump, 50 mm Hg for the Vienna pump and 125 mm Hg for the Nimbus pump.

The reason why the Tesla pump characteristic is different is not fully understood but may be a function of the action of the discs on the fluid. Wiseman, (1994), indicated that the pressure generated by a Tesla pump '*is inherently a product of its rotational speed and its runner diameter*'. This implies that the rotor peripheral velocity controls the pressure generated by the pump irrespective of the delivery. If this is correct, a larger diameter rotor running at a lower rotational speed but giving the same peripheral speed than the present pump should give the same pressure. This remains to be confirmed.

The Tesla pump constant pressure characteristic is confirmed by comparing the differential and delivery pressures over a range of speeds. Figure 6.8 shows a comparison of Tesla pump constant speed results with the delivery set at 10 l/min. The differential and delivery pressures relative to each speed are almost constant. The difference between the differential and delivery pressures at a delivery of 10 l/min at each speed is shown on Figure 6.9 which shows that the pressure difference reaches a maximum of less than 17 mm Hg at 1500 rev/min and approximately 13 mm Hg at 3500 rev/min. In pumping terms this difference is the negative pressure at the pump inlet.

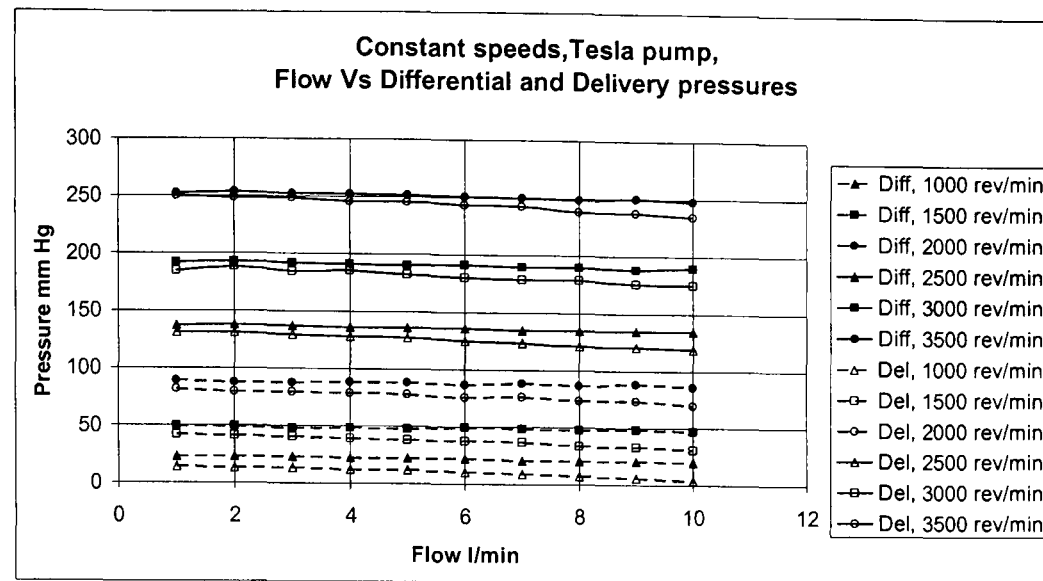


Figure 6.8. Speeds and pressures for the Tesla pump at constant speeds set at a delivery of 10 l/min.

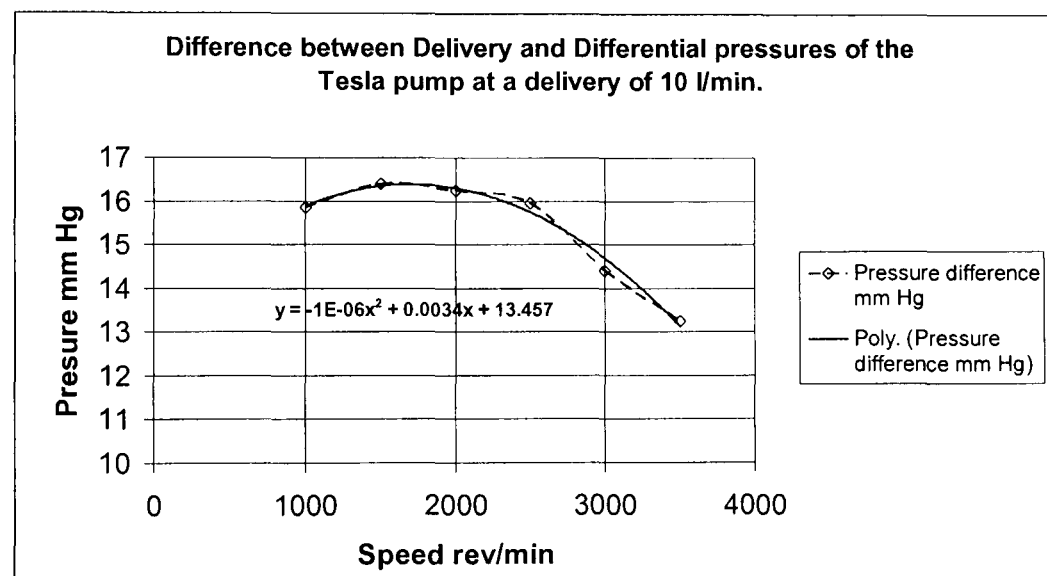


Figure 6.9. The pressure difference between delivery and differential pressures for the Tesla pump at a delivery of 10 l/min.

6.1.2 Non dimensional results.

The non dimensional results obtained from the Tesla pump have been compared with results from the paper by Crawford and Rice, (1974). This has proved difficult as the results from the paper are all calculated results with fixed parameters for the duration of the calculation. These results are therefore not strictly comparable as 'like for like'.

Three of the calculated results from Crawford and Rice, (1974) are shown on Figures 6.10, 6.11 and 6.12.

A sample of the Tesla pump results have been plotted using the same parameters as shown in the paper. These are shown in Figure 6.13. Note: - The values for U_o have been multiplied by 100 so that they can be seen on the same scales as the rest of the results.

Figure 6.10 shows the relationship between the calculated Reynolds number and efficiency with constant values for the through flow U_o . This indicates that maximum efficiency is reached at a Reynolds number of 5. In practical terms this is a very low figure as applying the formula to the Tesla pump gave typical Reynolds numbers in the range of hundreds. Enquiries were made from Arizona State University to obtain an explanation but were unsuccessful.

Using the formula quoted by Crawford and Rice, (1974), and assuming water as a fluid and a speed range of 1000 to 4000 rev/min., the gap between the discs was calculated for a range of Reynolds numbers from 1 to 10. The calculated gap was 0.3 to 0.05 mm. This is smaller than the gap(s) used in this thesis and is assumed to be the reason for the difference between the values of the Reynolds numbers. There is no information on gap sizes in the paper by Crawford and Rice, (1974).

The results on Figure 6.10 show that as through flow, U_o , becomes smaller, the efficiency increases. The combination of the Reynolds number and U_o results in a maximum efficiency of greater than 90%. This is a high but it is a calculated figure, does not take account of any physical losses and may not occur in practice.

The Tesla pump Reynolds no. Vs Efficiency result, shown on Figure 6.13 has similar shape to the calculated plot shown on Figure 6.10. The efficiency is low with low values of Reynolds number and rises to a peak value as the Reynolds number increases but then stays relatively constant up to a Reynolds number of greater than 600.

The values of the Tesla pump through flow parameter, U_o , at 0.01-0.03, are similar to the calculated values but are lower than the values of 0.02 to 0.5 quoted by Crawford and Rice, (1974).

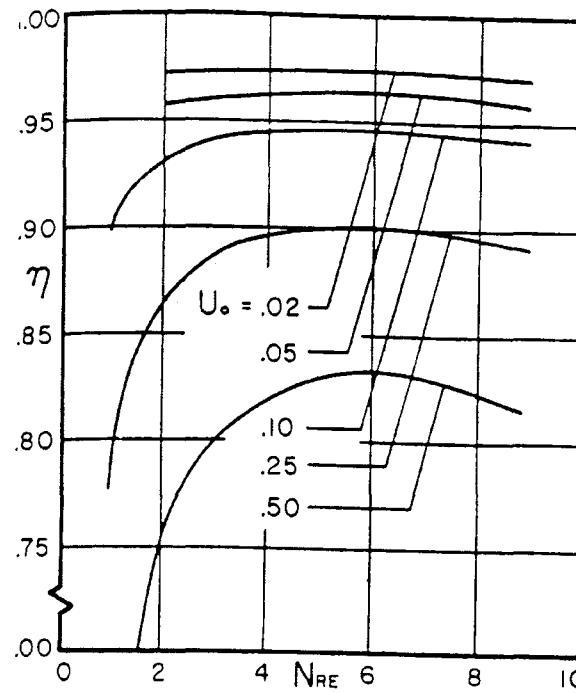


Fig. 1 Dependence of efficiency, η , on Reynolds number, N_{RE} , and on flow rate parameter, U_0 , with $r_0 = 5.0$ and $V_0 = 0$

Figure 6.10. Calculated Reynolds number and Efficiency with constant values for the through flow U_0 .

Figure 6.11 shows the calculated comparison of Reynolds number and pressure parameter with constant values of the through flow, U_0 . These curves have a similar shape to the previous result shown in Figure 6.10 as there is a peak value of the pressure parameter at a Reynolds number of 5 and that the value of the pressure parameter increases as U_0 decreases.

The Tesla pump results, Figure 6.13, show that the pressure parameter P_t , at a value of around 1 is relatively constant over the range of Reynolds numbers although it does not compare with the results from Crawford and Rice, (1974) at around 20 to 25 at the value of the through flow parameter U_0 of 0.01-0.03. In terms of curve shape this result is a good comparison with the calculated results shown on Figure 6.11.

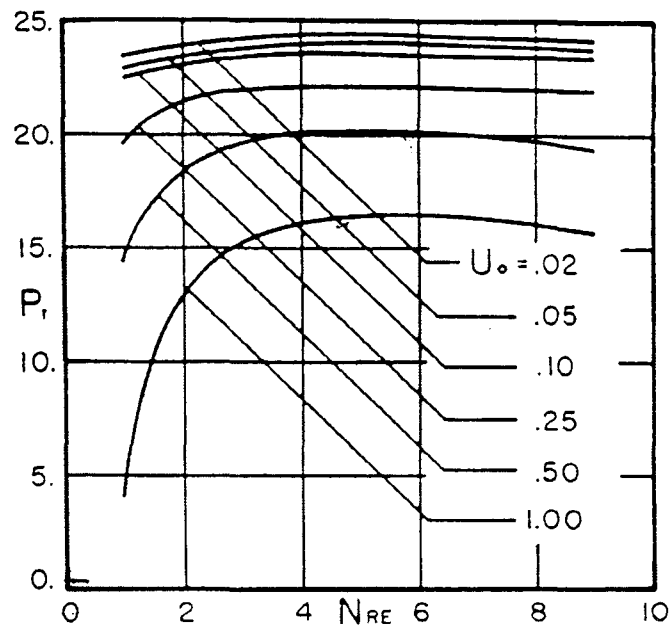


Fig. 3 Dependence of dimensionless stagnation pressure p_t , on Reynolds number, N_{RE} , and on flow rate parameter, U_o , with $r_o = 5.0$ and $V_c = 0$

Figure 6.11 shows the calculated comparison of Reynolds number and pressure parameter P_t with constant values for the through flow U_o .

Figure 6.12 shows the calculated result of Reynolds number and torque parameter T_1 with constant values of through flow U_o . This shows that for a given value of U_o , T_1 stays relatively constant over a wide range of Reynolds numbers and that T_1 increases as the value of U_o increases.

The Tesla pump Reynolds number Vs Torque parameter plot shown on Figure 6.13 is not the same shape as the calculated plot. The torque parameter has a high value at low values of Reynolds number and reduces as the Reynolds number increases but stays relatively constant from a Reynolds number of 300 upwards.

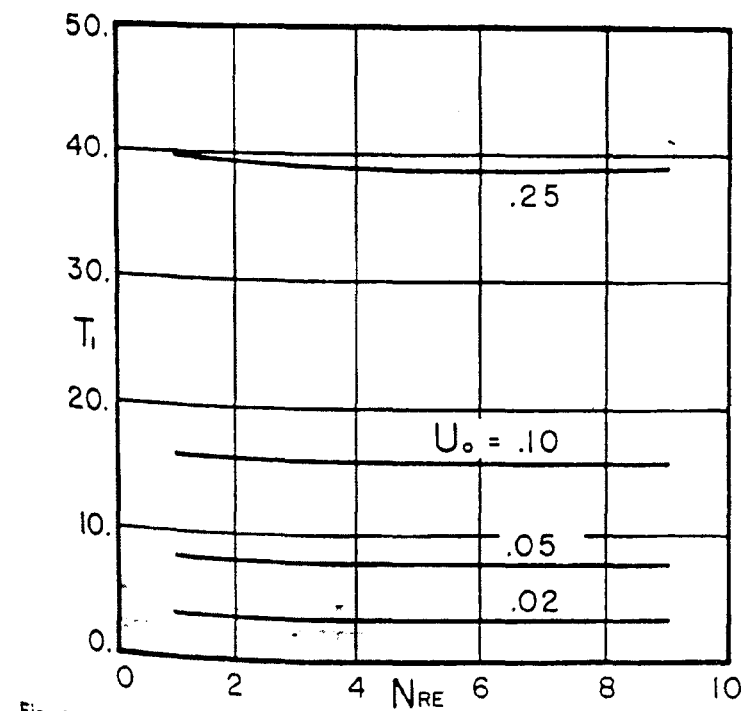


Fig. 2 Dependence of dimensionless torque, T_1 , on Reynolds number N_{RE} , and on flow rate parameter, U_o , with $r_o = 5.0$ and $V_o = 0$

Figure 6.12 shows the calculated comparison of Reynolds and torque parameter.

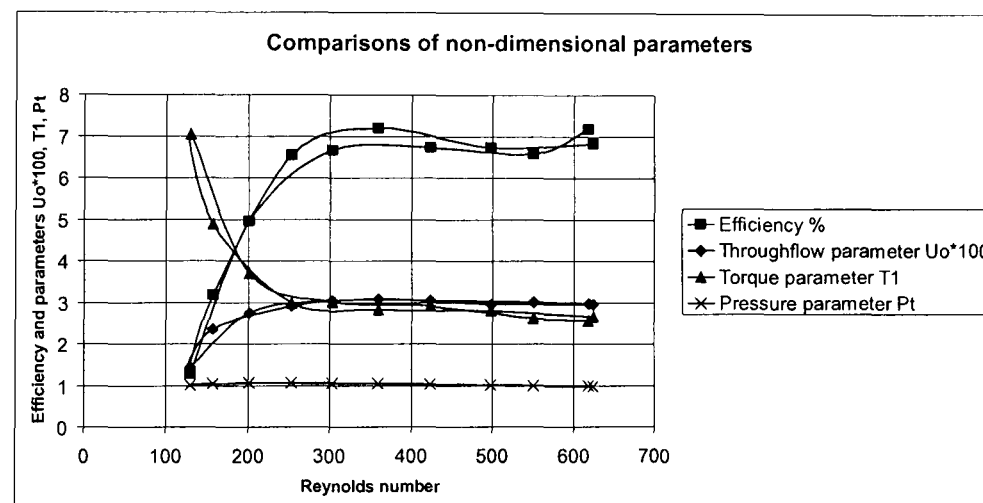


Figure 6.13. Tesla pump non dimensional parameters, Efficiency, %, Through flow parameter, U_o , Torque parameter, T_1 , and Pressure parameter, P_t , compared to Reynolds numbers, Re .

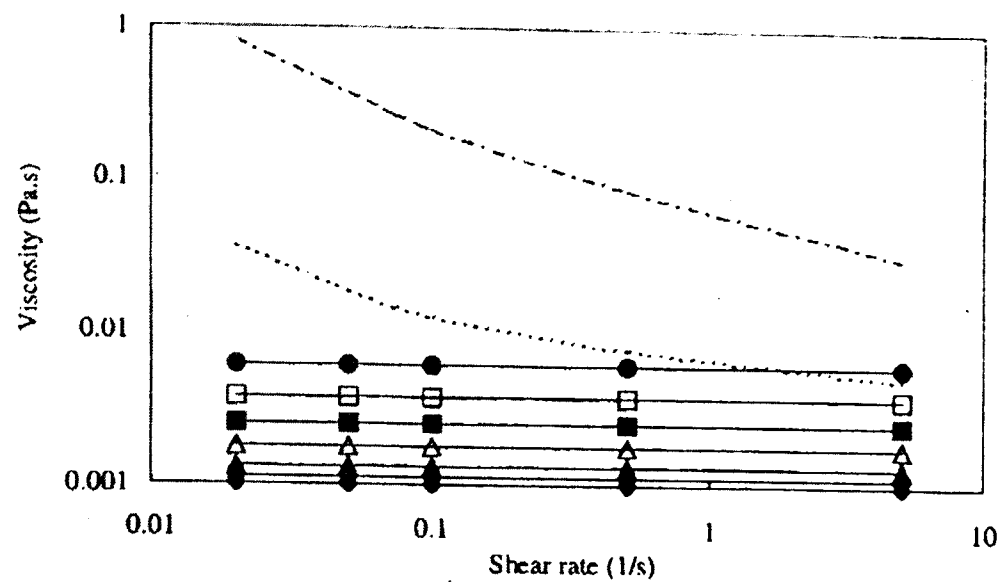
If the calculated results are to be believed, the Tesla pump tested here needs further work to improve the figures. Any possible improvements will not be discussed here.

6.1.3 Viscosity results.

Blood is a complex fluid as the viscosity changes relative to the shear rate making it a non Newtonian fluid. Attempts have been made by researchers to develop an analogue fluid to mimic blood but there does not seem to be a satisfactory substitute.

Figure 6.14 is a reproduction of one of the results from Wickramasinghe R. (2002). It shows the viscosity / shear relationship of water and various mixtures of water and glycerine and shows the viscosity is constant over a range of shear. These fluids are considered to be Newtonian. For the viscosity of water, a value of 0.001 Pa.s was assumed and it is so close to the other results it is not possible to show it clearly. (1 Pa.s is equivalent to 1 Ns/m² and viscosities will be expressed in Ns/m² in this thesis).

Two of the results in Figure 6.14 used mixtures of 40% and 50% glycerine. From previous work in this thesis, a mixture of 45%, by weight, glycerine in water had a viscosity similar to blood but without the viscosity / shear relationship of blood. From Figure 6.14, a line was calculated to represent a 45% glycerine / water mixture and this was used as a comparison to the Middlesex 45% glycerine / water result.



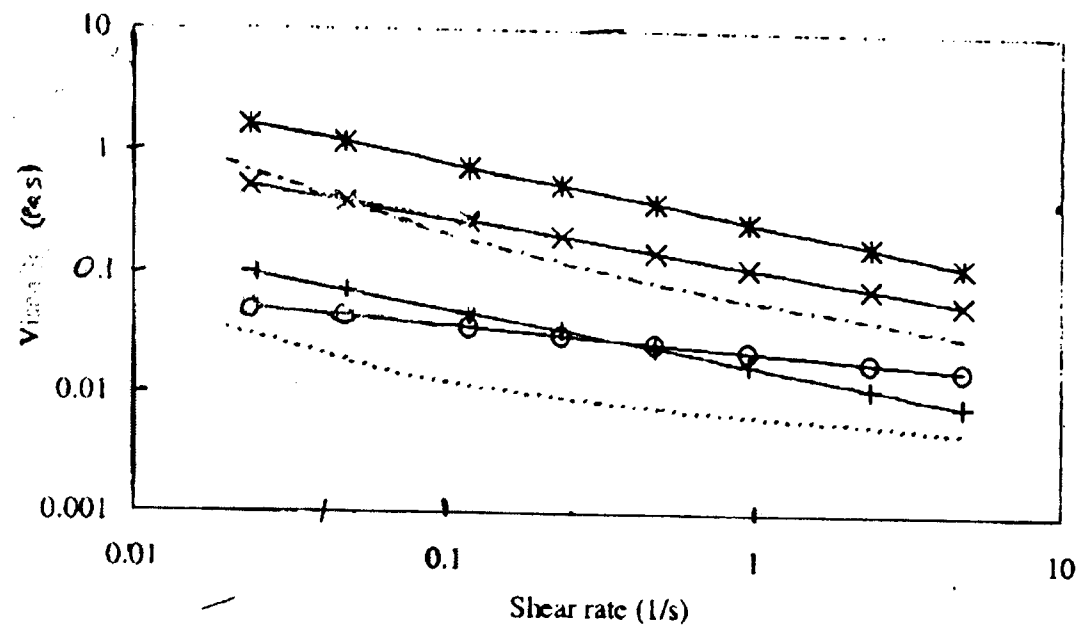
Symbol	% Water	% Glycerol	Symbol	% Water	% Glycerol
◆	100	0	◇	95	5
▲	90	10	△	80	20
■	70	30	□	60	40
●	50	50	25 % Hct blood	
			-.-.-.-.-	65 % Hct blood	

Figure 6.14. Results from Wickramasinghe S., (2002)
for mixtures of water / glycerine.

Figure 6.15 is a reproduction of the second of the results from Wickramasinghe R. (2002). It shows the mixtures of water / glycerine / xanthan gum tested for the relationship of viscosity against shear and is compared with pig blood at haematocrits of 25% and 65%. Human blood has an hematocrit of around 45% and why these results are not compared directly to human blood is not explained by Wickramasinghe. The blood results are

shown as dotted lines, the other results are shown as solid lines. It is noticeable that the results, although showing a change of viscosity relative to shear, are not curves in the same way as for pig blood. Inspection of the results for the water / glycerol / xanthan gum mixtures shows they are all straight lines and with the viscosity changes depending on the shear rate, these are all non-Newtonian fluids. The slope of these lines would indicate that any mixture of water / glycerol / xanthan gum would not be representative of blood over a range of shear although may be exact at a single value of shear depending on the concentration of the components in the mixture.

Two mixtures shown in Figure 6.15 appeared to be reasonably parallel to the blood viscosity / shear results over the shear rates shown. From Wickramasinghe one of these mixtures was shown to be 99.99% water, 0.01% xanthan gum. It was obvious that if the shear rate were to be greater than 10, that this mixture would not be representative of blood although it may be close to 45% hematocrit blood in the shear rate range of 0.1 to 1. The other mixture of 59.9925% water, 40% glycerine and 0.075% xanthan gum also may not be representative of blood above a shear rate of 10. Of the two mixtures the one most likely to be representative of 45% hematocrit blood in the shear range shown is the water / xanthan gum mixture.



Symbol	% Water	% Glycerol	% Xanthan gum
O	59.9925	40	0.0075
X	59.96	40	0.04
+	99.99	0	0.01
*	49.925	50	0.075
.....	25 % Hct blood	-----	65 % Hct blood

Figure 6.15. Results from Wickramasinghe, (2002) for mixtures of water / glycerine / xanthan gum.

From Appendix 2, the physiological wall shear rates vary from 43 to about 2000. Although the paper by Wickramasinghe, (2002), was concerned with blood gas transport, why were the blood viscosity / shear rate results published at shear rates different to the physiological rates?

Tests were conducted at Middlesex University using water and mixtures of water / glycerine and water / xanthan gum. It was not possible to test any other fluids due to a Campus move. Additional tests combining water, glycerine and xanthan gum may have provided a fluid that was close to blood in terms of viscosity at the particular shear rates considered.

The results of tests at Middlesex using water only are shown on Figure 6.16. The viscosity of 0.00094 is very close to the quoted value of viscosity of water from Rogers and Mayhew, (1983) of 0.001002. From this result, the methods used to determine viscosity were considered to be accurate enough for test purposes. Comparing them with the results on Figure 6.19 which is a comparison of results for all the mixtures tested it was not possible to distinguish results for water between the Middlesex, Wickramasinghe and Rogers and Mayhew, (1983).

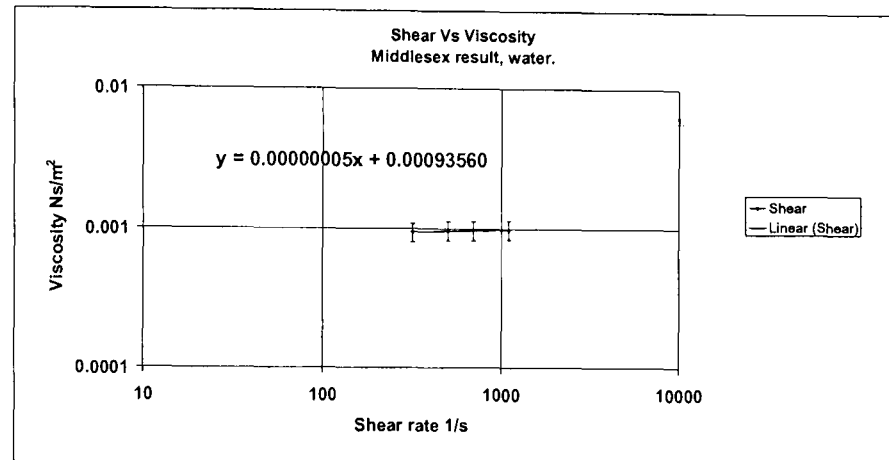


Figure 6.16. Middlesex result for water.

The Middlesex result for water / glycerine is shown on Figure 6.17. This gave a viscosity that was similar to blood, approximately 0.0037 N.s/m², but would only be comparable at one value of shear as the viscosity of blood changes as noted from Figure 6.19. The relationship of shear and viscosity of the water / glycerine mixture appeared to be a straight line with zero slope hence this mixture was assumed to be a Newtonian fluid and not representative of blood. The comparison of the Middlesex and Wickramasinghe results show little or no difference as shown on Figure 6.19.

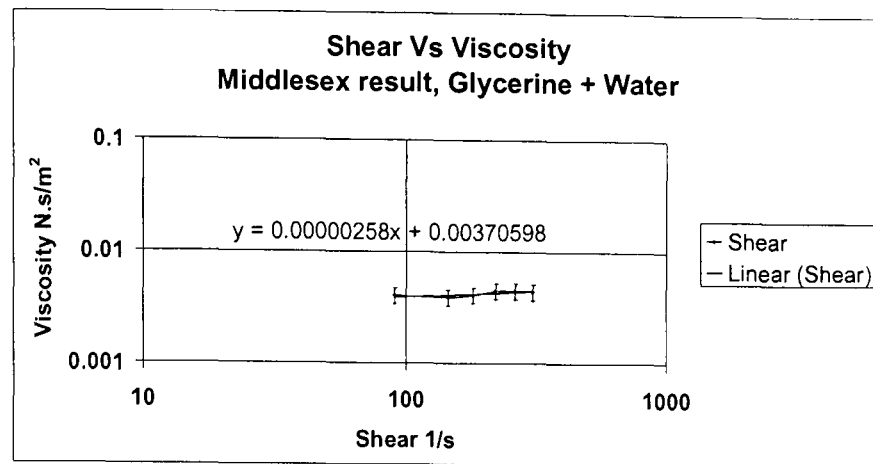


Figure 6.17. Middlesex result for water + glycerine.

The Middlesex result for water + xanthan gum, shown on Figure 6.18, shows that the characteristic had a slope that was effectively zero and it is not understood how the Wickramasinghe result was obtained as the Middlesex result is little different from water.

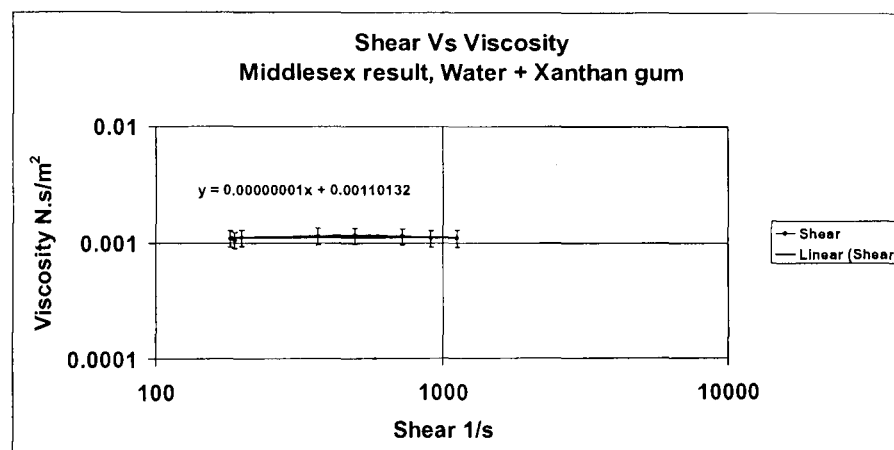


Figure 6.18. Middlesex result for water + xanthan gum.

Figure 6.19 is an attempt to compare all the viscosity results from Middlesex University and Wickramasinghe, (2002). This plot is obtained by taking the results obtained by the tests conducted at Middlesex University and extrapolating them to cover the same shear range as used by Wickramasinghe. Figure 6.19 shows that the Middlesex results for water / xanthan gum do not compare favourably with the results for Wickramasinghe. The reason for this is not understood. The mixture ratio of the water and gum was checked and found correct according to the paper. From

Wickramasinghe, the Middlesex xanthan gum result should have shown a decreasing value for viscosity with increasing shear rate as shown on Figure 6.15.

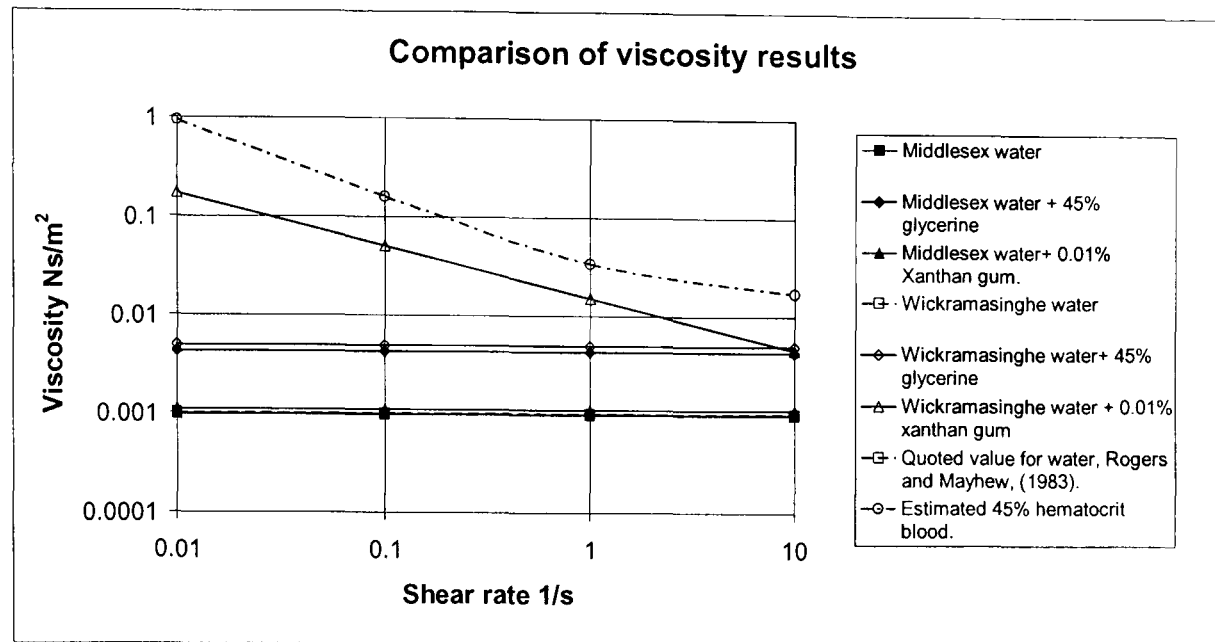


Figure 6.19 Comparison of results for all the mixtures tested.

Brookshier and Tarbell, (1991) found pig blood provided 'a good match of the complex viscosity of human blood over a wide range of hematocrits, (20-79 %) and shear stress, ($1-1000\text{s}^{-1}$). Brookshier and Tarbell, (1993), attempted to produce a transparent, stable, shear thinning fluid that was a close match to human blood. Their results, a mixture of 0.04% xanthan gum + 40% glycerine + water, appeared to give a reasonable result compared to 46% hematocrit pig blood over a shear range of 1 to 1000. This fluid may be a good starting point for any further tests

They concluded that 'the elastic properties of the fluid did not play a significant role in the flow behaviour'. This implies that the performance of a blood pump would not be affected by the elastic nature of blood.

6.1.4 Cavitation.

The Tesla pump does not appear to have a problem with cavitation. Tests have been run with a restricted inlet resulting in the same performance as an open inlet.

Figure 6.20 shows the flow Vs speed comparison of two tests, one with an open inlet, the other with a restricted inlet, for 10 l/min @ 200 mm Hg.

There is little or no difference in performance between the two settings. A similar result is shown on Figure 6.21 for the flow Vs differential pressure at the same set point.

The flow Vs speed and flow Vs pressure differential tests were repeated for 5 l/min @ 100 mm Hg at the same valve settings, as shown on Figures 6.22 and 6. 23 respectively. Again there was little or no difference in results.

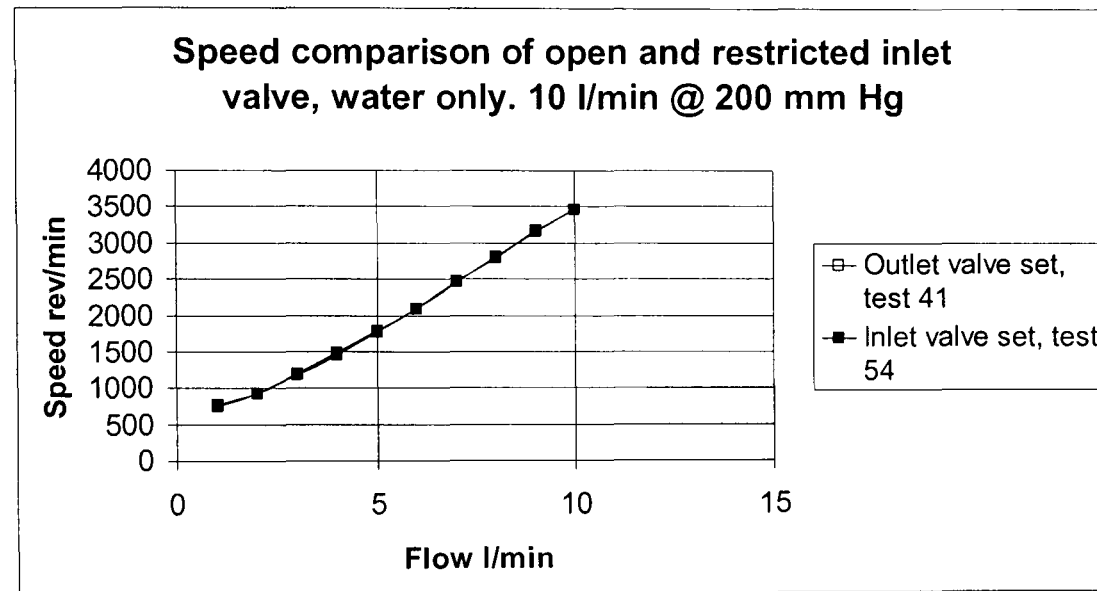


Figure 6.20. Speed Vs flow results comparing the outlet valve Vs the inlet valve to control the delivery at 10 l/min @ 200 mm Hg.

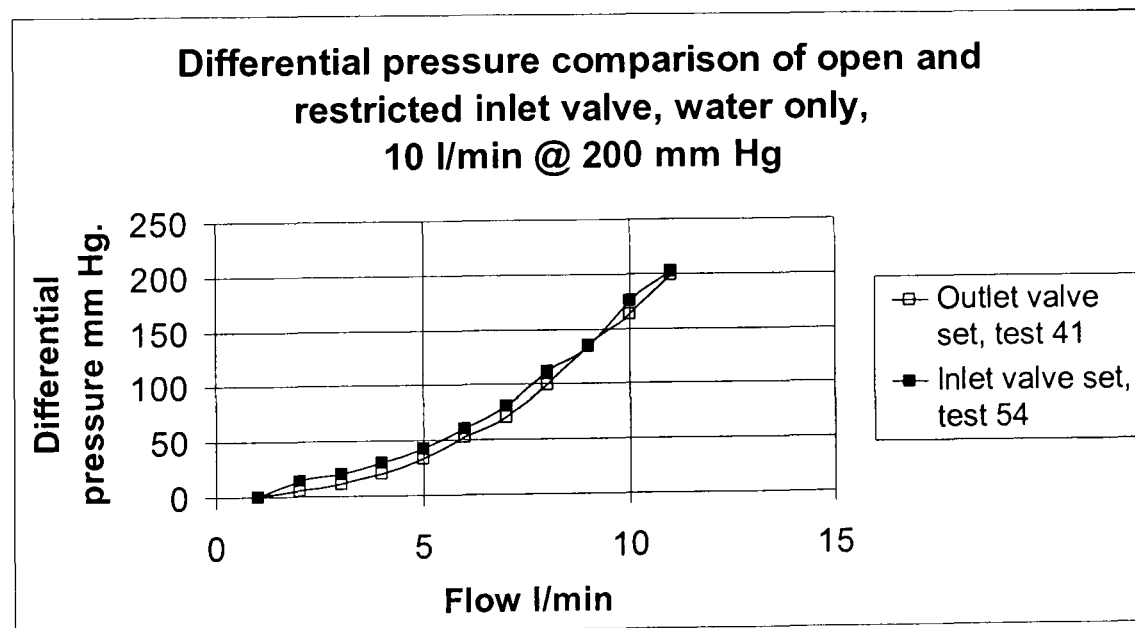


Figure 6.21. Differential pressure Vs flow results comparing the outlet valve Vs the inlet valve to control the delivery at 10 l/min @ 200 mm Hg

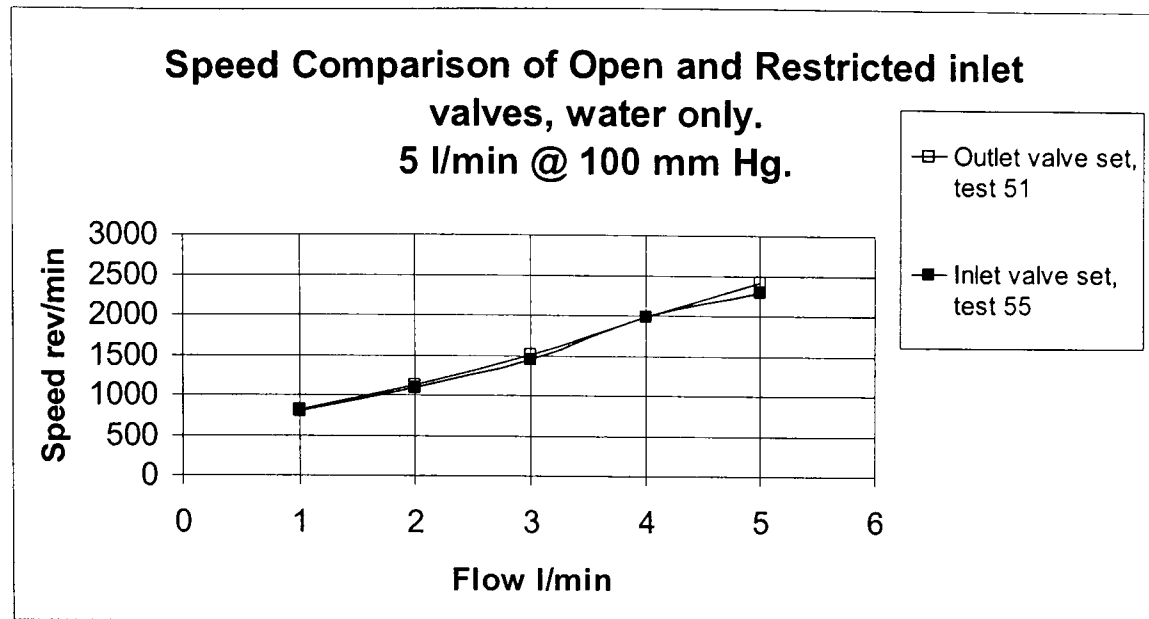


Figure 6.22. Speed Vs flow results comparing the outlet valve Vs the inlet valve to control the delivery at 5 l/min @ 100 mm Hg.

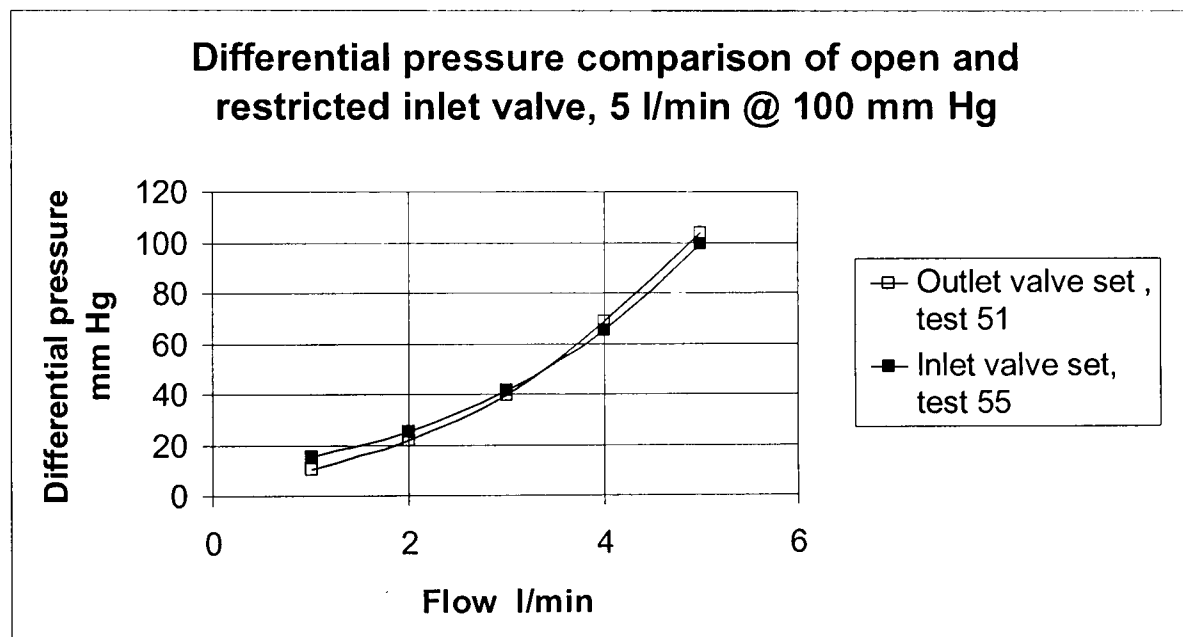


Figure 6.23. Differential pressure Vs flow results comparing the outlet valve Vs the inlet valve to control the delivery at 5 l/min @ 100 mm Hg

There were no physical signs of cavitation during any of these tests. The pump delivery was stable during the tests and there were no additional noises or bubbles noted in the output. It is not known if other blood pumps have been tested for this condition as no evidence has been found of any tests to induce cavitation.

Barske, (1960) tested a Tesla type pump to promote cavitation. It was tested over 'long test runs at speeds of up to 18,000 rev/min and inlet pressures throttled down to 1 lb/in² abs., (52 mm Hg), but no cavitation marks

were produced on its surfaces'. A discussion contribution to the paper by R. Sadek offered a design of impeller that in his opinion '*was cavitation free*' but no results or other information was offered.

6.1.5 Implantation.

The installed volume is not always listed but some of these pumps are at least equal in size to the human heart. There are no reports in the literature to show the installation of pulsating pumps in terms of creating the volume inside the body or how the pump is fixed in position to prevent movement or strain on the connections to the blood circuits. This must be of concern as pulsating pumps are likely to move as the pulsating action takes place.

Radial and axial flow pumps tend to be smaller and lighter than the pulsating types. This will make installation more acceptable as less room will be needed with less chance of damage to the body. As these pumps do not pulsate, the chances of pump movement are reduced.

The problem of heat generation appears to be negligible due to a combination of good pump designs that do not generate excessive heat and adequate blood flow to conduct the remaining heat away that is then dissipated throughout the rest of the body.

The Tesla pump was tested for temperature rise, Figure 6.24. The pump was set to deliver 1 l/min at a pressure of 200 mm Hg and ran for 5½ hours. A temperature rise of 16° C was recorded from an ambient temperature of 20 -21°C. This gave an average temperature rise of 0.048°C / min over the test period. There was no cooling supplied for the pump and the output of the pump was fed straight back into the inlet.

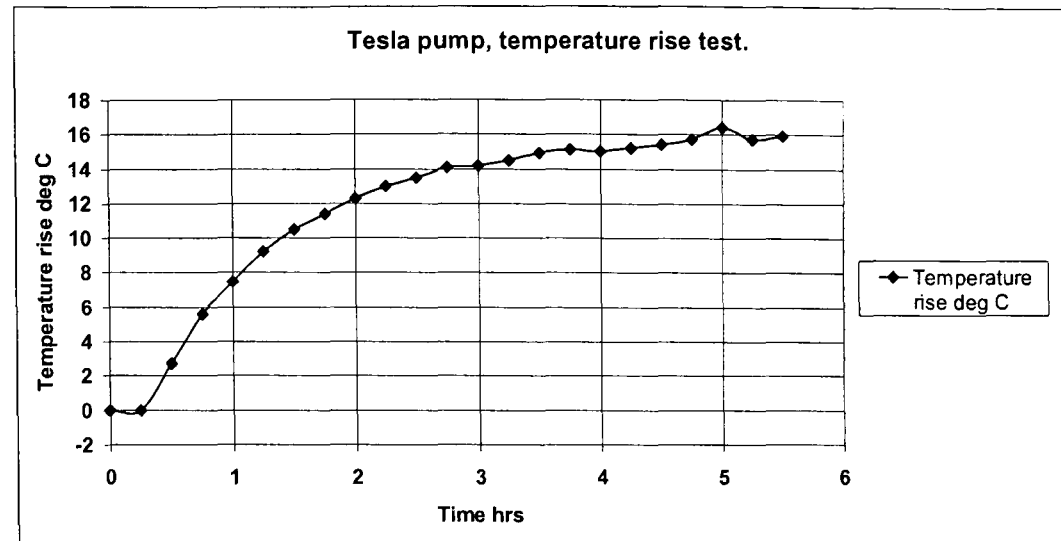


Figure 6.24. Tesla pump fluid temperature rise test.

Takami et al., (1996), using a pivot bearing support pump to evaluate efficiency and heat generation, recorded blood temperature rises of up to 16°C in 4 minutes. If these results are correct, the Tesla pump should be satisfactory as the rate of temperature rise is lower.

Control systems have been developed that can be implanted together with the pump. These systems must not have excessive temperature rises. The temperature limit of 41°C, discussed previously, is important. During tests on an implanted pump by Okamoto et al., (2006), the implanted electronic control unit was kept at a temperature under 45°C but examination after the tests showed burns on the tissue surrounding the unit.

The effect of temperature on muscle performance has been investigated. As the temperature rises to around 40 ° C, skeletal muscle tends to exhaust sooner compared to muscle at normal temperatures, (~37°C), as shown by Walters et al, (2000), Nybo et al, (2001), Todd et al., (2005) and Thomas et al, (2006). No information has been found on the effect of blood temperature on cardiac muscle. If a blood pump raises the blood temperature sufficiently and the same effect applies, a damaged heart may be weakened further.

With the combination of the new power transfer systems and the newer pumps, it is now possible to completely implant a pump with the closure of the chest. There has been no attempt to make the Tesla type pump in this thesis suitable for implantation but only to investigate the potential of this pump type. Implantation may be possible for this pump after development.

6.1.6 Surface contact area.

In general, the smaller the pump, the smaller the blood contact area. With a pulsating pump, it is not possible to make the pump smaller, reducing the internal area, and maintain the blood flow rate without increasing the pump 'beat' rate and blood stress levels. This is a disadvantage with a pulsating pump.

Radial and axial flow pumps are generally smaller than a pulsating pump and the blood flow rate can be increased by increasing the pump speed as long as the blood stress levels are not exceeded.

Reducing the internal surface contact area, the use of biocompatible materials and surface conditions will be important to prevent blood clots forming. The use of metals and plastics such as titanium, the polycarbonates and polyurethanes are well established as well as the techniques for making biocompatible surfaces as developed for the HeartMate pump.

The Tesla pump type will have a large internal surface area due to the number of discs and this pump will be at a disadvantage considering blood reactions. Calculations indicate a rotor contact area of 0.089 m² for rotor 3 and 0.157 m² for rotor 7. The surface area of the volute has not been considered as it is a constant irrespective which rotor build is used. The difference in rotor surface contact area of rotor 7 of approximately 2 times compared to rotor 3 does not account for the difference in performance and without accurate information on the dimensions for any of the other pumps discussed, no comparison of surface areas can be made.

Tests by Pierrat and Nassoy, (2004), to detach red blood cells from selected surfaces gave forces ranging from 1.4^{-9} N to 0.45^{-9} N depending on the surface used. The results of similar work by Bowers, et al., (1989), indicate that forces ranging from 6.29^{-9} N to 3.06^{-8} N are necessary to remove red blood cells from selected surfaces.

The tangential forces generated with the Tesla pump gave figures from $4.12 \cdot 10^{-25}$ N to $1.32 \cdot 10^{-23}$ N depending on radius and speed. It is therefore doubtful that the forces generated will remove clotted blood from the surface of the discs.

6.1.7 Systemic anticoagulation.

Systemic anticoagulation is where anticoagulants such as heparin are added to blood to prevent blood clots forming. Blood clotting can be provoked for two reasons, a) surface reactions and b) excessive blood stress levels. If the materials and surface finishes are suitable there should be little or no problems with surface reactions and this has been covered previously.

The running of the pump could cause blood stress problems however.

Reul and Akdis, (2000), indicate that there are two problems concerning stress levels with rotary pumps;

a) The circumferential tip velocity of the rotor should be limited to a maximum of around 10 m/s otherwise problems with cavitation and maximum allowable blood shear rates can occur.

Applying this to the Tesla pump design, limiting the peripheral speed to 10 m/s at a rotor diameter of 60 mm, a maximum rotational speed of around 3200 rev/min is indicated. With the present Tesla pump this will give a delivery of 4 to 10 l/min at this speed depending on the set point and pump construction. The output can be varied by changing the number of discs on the rotor. More discs are expected to give a higher delivery for the same pressure at the same rotational speed.

b) In most commercial blood pumps Newtonian shear stress is in the range of 20-100 N/m² and axial pumps can reach around 400 N/m².

For the present Tesla pump, a shear stress of ~ 30 N/m² has been calculated at a speed of 4000 rev/min. This figure is still high when white blood cells appear to have a maximum stress of 15 N/m² before damage to the cell.

Lowering the pump speed and / or changing the pump construction will reduce stress levels but also reduce pump delivery. With the present pump the speed for this stress is 2050 rev/min, as shown on Figure 5.17. This will give a delivery of 2.5 to 7.5 l/min depending on set point and pump construction.

The first generation blood pumps imitated the biological heart. For these pumps, valves were necessary to control blood flow direction and have been a source of problems. Many valve types have been made and range from 'ball in cage' types to flap valves made from biocompatible materials, animal and human tissue. There does not seem to be one type that is ideal and trouble free.

A pump without valves is likely to be a better pump as problems with valves are avoided. As rotary pumps and the Tesla pump do not use valves there will be no problems with this part of the application and it will not be considered further. However, a rotary pump will have to maintain a minimum speed to avoid regurgitation.

One of the many indications for heart transplant is a low systolic blood pressure; Pepper, (2000), indicates 80-90 mm Hg. The work by the blood pump will stress the blood but as long as the stress level is equal to or less than the biological levels there should be no appreciable blood damage. The Tesla pump appears to produce low blood stresses of less than 30 N/m² with the present construction. Changes to the disc design and assembly of the pump indicate that the stress levels can be reduced further although the effects on the pump performance are not known at this point. Following these results the Tesla pump would seem to be more suitable to pump blood than some of the other pumps described in this thesis.

6.1.8 Simple mechanical blood propulsion system.

The second generation pumps were rotary pumps and used one moving part, the impeller. This pump type was radial flow or axial flow. The first axial flow pumps used shaft drives powered from outside the body and seals on the drive shafts to keep the blood inside the pump. This is not a successful method and problems were experienced with seal leakage and blood damage due to heat generation at the seal face. Efforts to solve these problems with flushing systems to cool the pump meant that this type of pump became complicated and has now been superseded.

Developments with power transfer systems have produced a system that transfers enough power through the skin to operate an implanted blood pump without damage to the skin. This makes the pump completely implanted with no risk of infection through power cables or pipes passing through the skin. Some rotary pumps use blood immersed bearings. An example of this type is the Jarvik 2000, an axial flow pump. The bearings are small in diameter in order to keep the rubbing speeds down. Westaby, (2000), quotes that the Jarvik 2000 uses '*blood immersed bearings 1 mm in diameter*'. These bearings are satisfactory for this application because they are lightly loaded and are made from biocompatible materials. Other pumps use jewelled or ceramic bearings for durability and biocompatibility. The amount of blood damage does not appear to be excessive and is assumed to be within the range that the body can handle without severe reaction.

The third generation blood pumps use magnetic or fluid force bearings to position the rotor. In this design the rotor is completely suspended and officially does not touch the inside of the pump casing. The Incor Berlin axial flow pump and the HeartMate III rotary pump are both examples of this design.

The InCor Berlin and HeartMate III pumps are examples of pumps with no physical bearings with the rotors magnetically suspended and driven. The position of the rotor in the InCor Berlin pump is sensed and is used to calculate the blood flow rate. The rotor speed is adjusted to maintain the correct flow rate. A flow rate calculation system does not appear to be used on the HeartMate III or any other of the rotary pumps and they do not appear to be designed for this.

With the HeartMate III, the pump can be separated into two parts, the blood pumping section and the drive and electronics section. With this construction the blood contacting part of the pump can be changed independently of the other. This makes the electronics section reusable and the blood contacting part single use and disposable.

The present Tesla pump design uses a shaft drive with seals but there is no reason why a redesign could not produce a pump with the pumping element mounted on blood immersed bearings and magnetically driven from outside the pump casing. This would give a pump with only one moving part as proposed by Westaby, (2000).

6.1.9 System reliability.

The control system of the pump should be as simple as possible for the human operator, even to the extent of not needing any human intervention. It may be necessary to compute speed or blood flow and this should be possible with a microprocessor system using feedback control to keep the pump delivery within acceptable limits. Any external controls should be as few as possible and any natural characteristics of the system should be utilised. The electronic controller for a magnetically levitated pump will be more complicated compared to a pump with mechanical bearings as it will have to maintain blood flow and control the position of the rotor. This may lead to more errors and a greater possibility of failure.

In terms of systems reliability, any blood pumping system cannot be 'fail safe' if the pump is using only one system, because if that system fails, the pump stops. If complete reliability is required, then multiple control

systems are needed. If the primary system fails, another takes over automatically without interruption and signals that a system failure has occurred. This will give security to the patient and time to repair or replace the faulty system. The system will have to be a minimum of two levels but a three level system would be preferred. Whether this can be done in terms of cost and complexity needs discussion.

6.1.10 Delivery.

The requirement for adult patients as indicated by researchers is about 5 litres/min at a pressure of 100 mm Hg. Some of the pumps described in this thesis deliver more, others less than this figure. It is assumed that those pumps that deliver less than 5 litres/min are used for organ perfusion support and not full cardiac assistance.

There does not appear to be a 'standard' specification for delivery that is accepted by the industry that all researchers work towards. This makes comparisons between various pump designs difficult.

6.2 Comparison with other pumps.

6.2.1 Size and output

The present Tesla pump is capable of delivering 10 l/min at a differential pressure of 200 mm Hg. at less than 4000 rev/min. This delivery is in excess of what is required in terms of biological support which is usually quoted as 5 l/min at an afterload pressure of 100 mm Hg. As an example, the HeartMate III pump is stated to deliver 7 l/min at 135 mmHg for a speed of 4800 rev/min., Schöb and Loree, (2005), (This is a rotary pump with a magnetically suspended bladed rotor). This comparison indicates that the Tesla pump is too large for the application and that a smaller pump would be more suitable.

The physical sizes of the pumps are difficult to compare and the figures quoted on Table 2.10 are an approximation.

6.2.2 Pump configuration.

The present Tesla pump design as shown in this thesis is not suitable for a blood pumping application. It is shaft driven and a seal is used to prevent fluid leakage along the shaft. Seals have been mentioned previously as a source of wear and heat generation problems. The miniature ball bearings on the shaft would not be acceptable for a blood pump application due to the need for lubrication.

This design was considered to be suitable only to determine if a Tesla type pump could be used to pump fluids with characteristics approximating to blood.

The HeartMate III and other pumps indicate that the rotary pump concept works, although more information is required in terms of pump endurance and subsequent blood condition. A lack of long term performance results *in vivo* is a problem with all blood pumps, as most are removed as soon as the patient has recovered enough to do without the pump. The implantation time can be hours to days.

6.2.3 The general requirements of a blood pump.

The pump will need to be as small as possible in order that implantation inside the patient is possible without undue discomfort to the patient and to reduce the internal blood contact area. Due to the small size, the pump will have to run at high speed to deliver the quantity of blood. Careful design of the pump flow passages is necessary to provide smooth blood flow without excessive fluid stresses. This may lead to less anticoagulation procedures and previous chapters have outlined the problems with blood stresses and fluid flow problems.

The materials of the pump must be biocompatible. This implies titanium for the construction of the pump.

Consideration will have to be given to the assembly of the pump in terms of the relative positions of the inlet and outlet ports and the anatomical requirements of the patient. The connections to the patient must be acceptable and, if they exist, standard connections must be used.

More work is required with the Tesla pump hydraulic performance to optimise the rotor configuration in terms of disc spacing and the disc inner and outer diameter ratios in order to determine a 'best' rotor for this application. It could be that a 'best' rotor is not possible following the present design of an assembly of same sized discs rotating on one centre line. Discs of different diameters with the largest disc in the centre of the rotor may be possible with the smaller discs acting as 'feeders' to the larger discs. This may produce a less abrupt change of stress and direction for the blood as it enters the pump.

6.3 Design suggestions.

Consideration will have to be given to evaluate the useful life of the pump together with the installation and power supply.

Assuming the pump does not have an indefinite life, or has a life shorter than a patient will need to use it, consideration to a safe means of changing the pump will be required. As far as possible, major surgery should be avoided when changing the pump. This may involve developing 'reusable' methods of mounting the pump 'to' the patient.

Such methods may involve the development of 'plug in' electrical connectors and 'plug in' blood connectors and pipe work.

It is hoped that a pump can be developed to fulfil all the biological requirements and patient expectations in regard to comfort.

6.4 Proposed new design of pump.

Before development work is started, the application must be clearly stated, for instance, the pump will be for use outside the body. This implies that size may not be particularly important but that secure fixings will have to be provided.

6.4.1 Disc construction.

It is proposed that the disc(s) comprising the rotor will be built with steel or other magnetic material as a spine. The discs will have to be biocompatible and not have any surface imperfections that could cause blood clotting problems. Depending on the forces required to drive the pump, it may be possible to build the rotor assembly with the end discs as the only magnetically attractive part of the rotor. The rotor will carry a spindle that will act as bearing shaft.

The aim of a redesigned rotor assembly is to optimise fluid flow between the discs and to take advantage of more efficient disc diameter ratios as it is suspected that the ratio used in the tests was not an optimum figure. The change to the diameter ratio can be achieved in two ways and these are outlined below.

Figure 6.25 shows the centre holes in the disc pack to be a gradually increasing diameter the nearer the fluid entry. This will create the impression of a 'tapered' entry and is similar to the methods employed in multiple branch pipe systems.

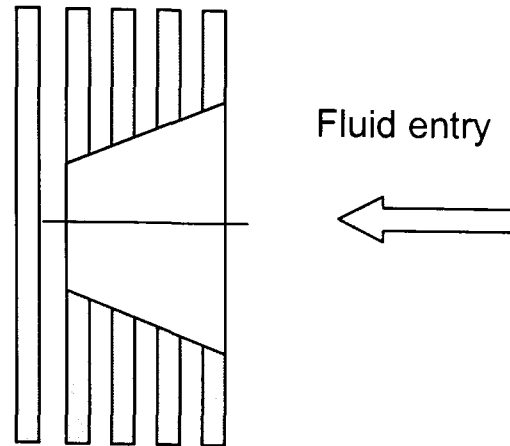


Figure 6.25 Rotor with 'tapered' central hole.

Figure 6.26 shows discs of various outer diameters, with the largest diameter disc in the centre of the pack and the smallest discs to be at the ends of the pack. The central hole to be tapered with the largest diameter at the fluid entry position. This may cause problems with the design of the volute passageway and cutwater.

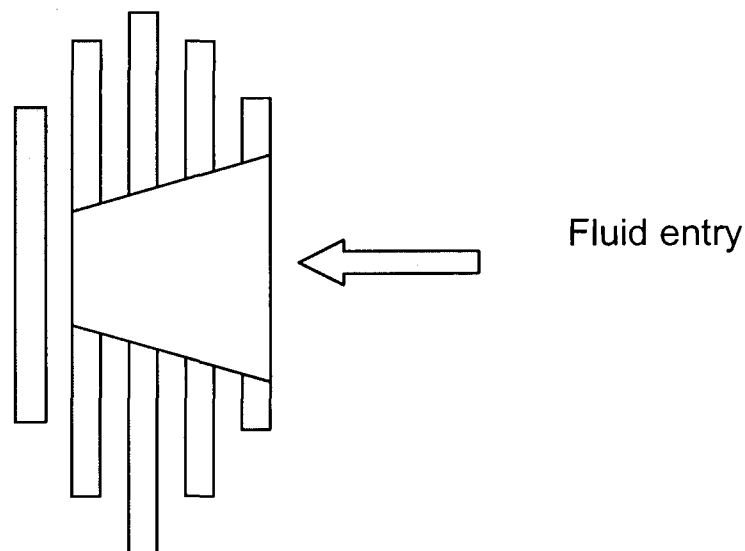


Figure 6.26 Rotor with various diameter discs and tapered central hole.

Both these configurations will give discs of differing diameter ratios with Figure 6.26 giving the greatest variation. If there is an optimum ratio of the disc outer diameter and the diameter of the hole through the disc, these

arrangements are more likely to find and operate at the most efficient point than discs with fixed sizes for the outer and inner diameters.

An important consideration for the differing diameter disc rotor is the variation in shear for a given rotational speed across the rotor. As shear is a product of linear velocity and the gap between the moving and stationary parts of a system, a large diameter disc running further away from the wall of the volute will produce a lower shear than a large diameter disc running close to the volute wall. A smaller diameter disc running at the same rotational speed but closer to the wall can produce the same shear depending on the disc diameter and gap. It may therefore be possible to produce a disc pump with a low shear figure across the rotor but with an improved performance.

The cross section of the discs should also be considered for flow improvements as the blunt end of the disc may lead to flow problems. Figure 6.27 shows a half section outline of an improved disc assembly. Each disc has a half tapered inlet at the central hole and tapers up to the periphery of the disc. The taper should help to reduce the stress levels of the fluid passing between the discs and between the end discs and the volute walls. The tapered design should also reduce the mechanical stresses imposed on the disc(s) as it rotates.

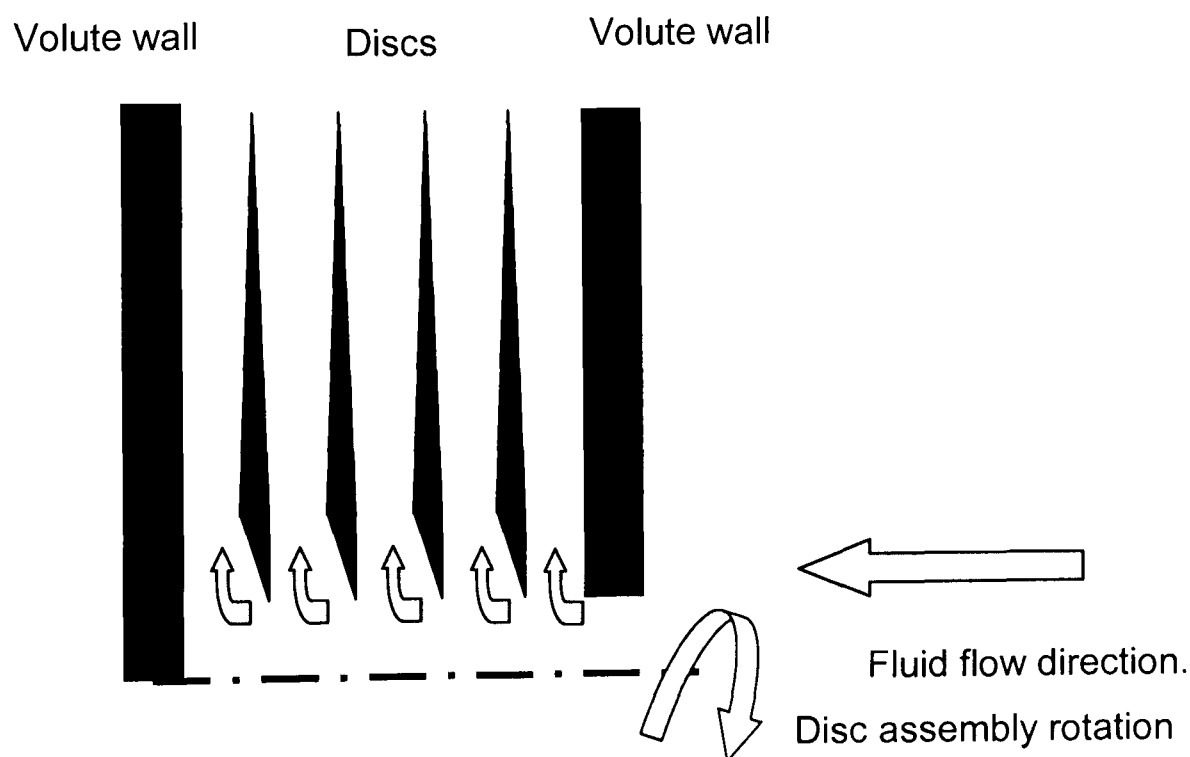


Figure 6.27 Half section through rotor showing proposed disc design.

Tapering the disc to give a 0.5 mm land at the outer edge of the first disc will leave a gap between the volute wall and the periphery of the first disc of 2.75 mm. Figure 6.28 details the disc design. Keeping the gap between the volute wall and the disc at the fluid entry point at 1.5 mm, at a speed of 4000 rev/min, the stress at the outer edge or periphery of the disc is reduced to 16 N/m². The stress for the untapered disc is 29 N/m². Changing the speed to 2000 rev/min, the stress reduces to 8 N/m² at the periphery of the tapered disc. This is very close to biological levels for maximum blood stress quoted on Appendix 2, as less than 6 N/m².

Figures 6.29 and 6.30 show the calculated results for stress Vs wall gap at the two speeds quoted.

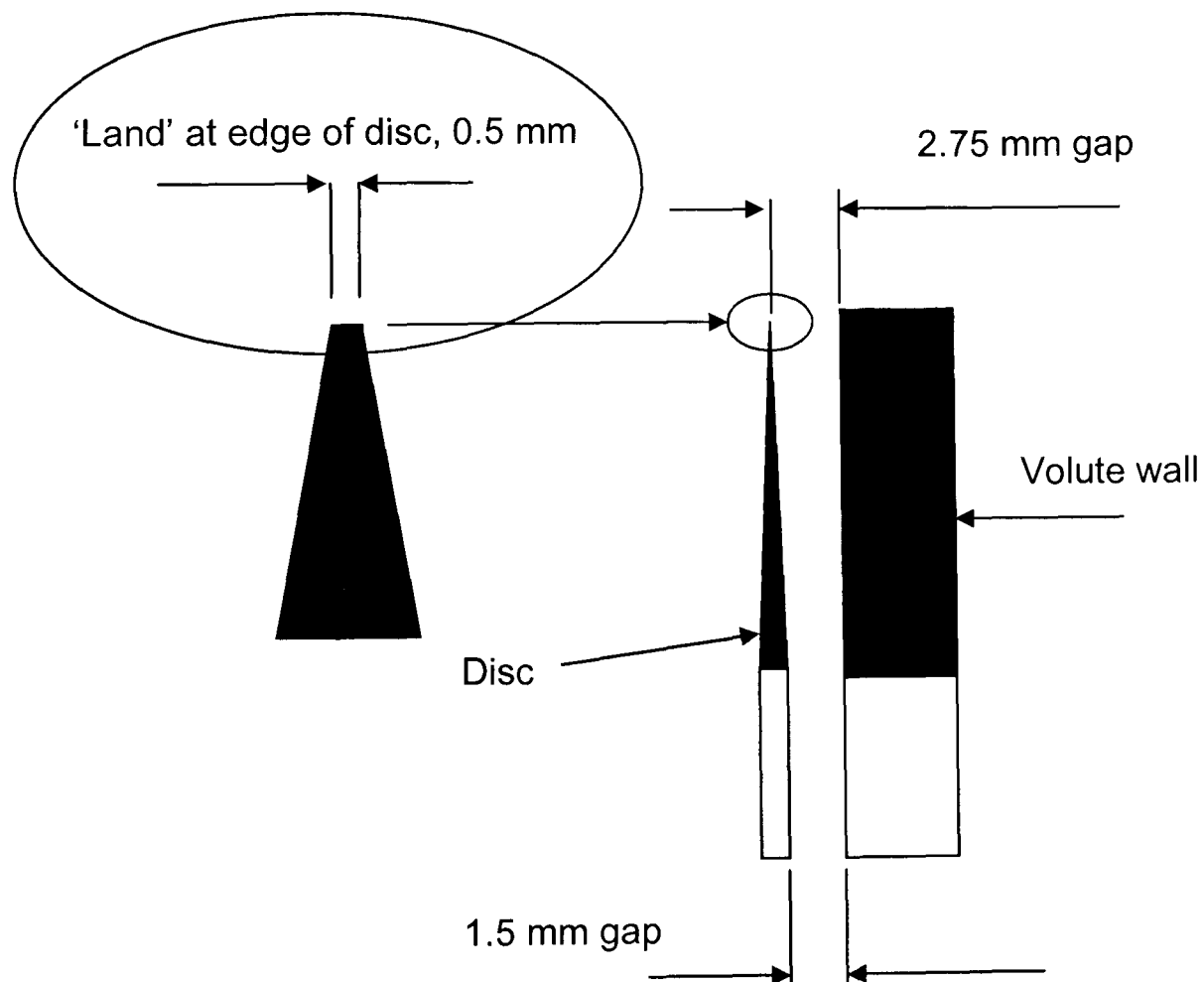


Figure 6.28. Details of the disc edge and calculated gaps.

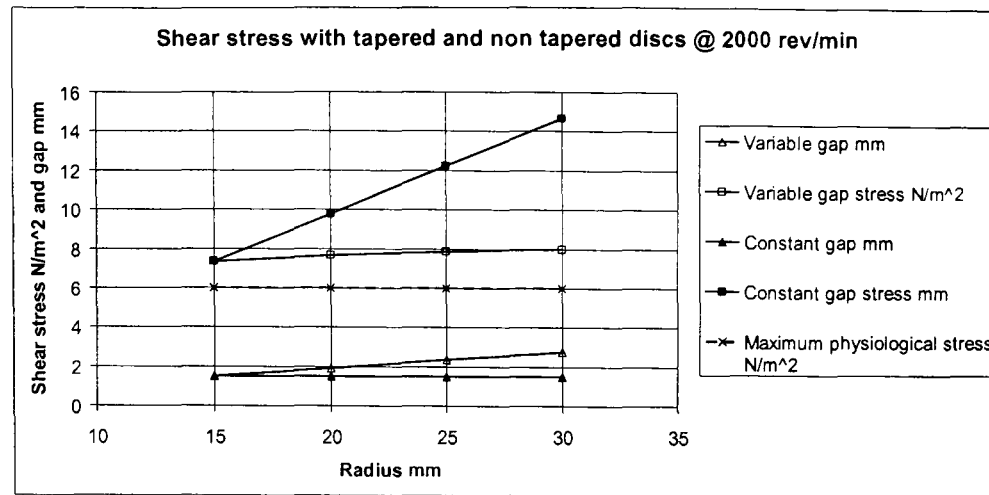


Figure 6.29. Calculated shear stress on tapered and non tapered discs at 2000 rev/min.

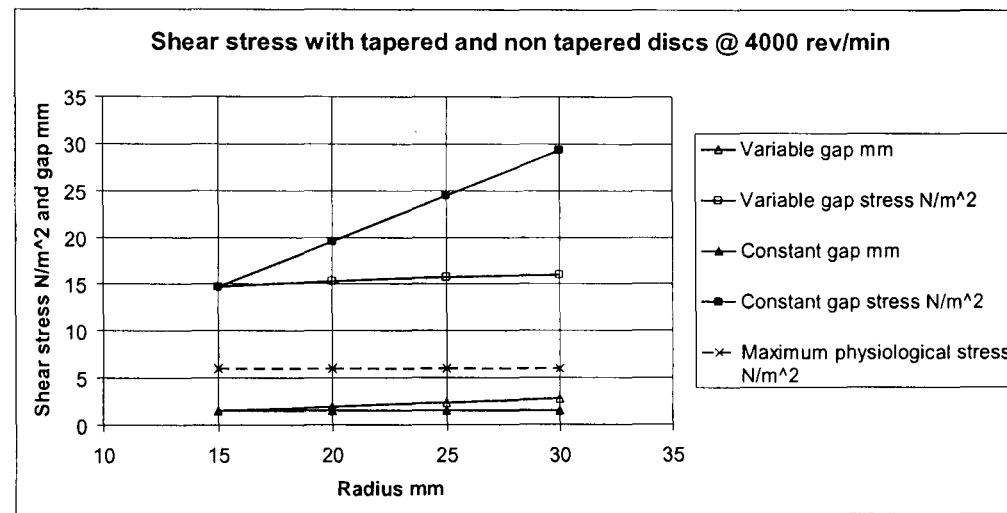


Figure 6.30. Calculated shear stress on tapered and non tapered discs at 4000 rev/min.

Assuming that a pump can be built that will give a constant shear stress across the gap between the first disc and the volute wall, calculations have been made to determine the size of the rotor / volute gap for a shear stress of 5 N/m^2 . This figure is below the maximum figure quoted on Appendix 2, Physiological Flow Parameters.

The result is shown on Figure 6.31 for pump speeds of 2000 and 4000 rev/min. The effect on pump performance with these gaps is unknown.

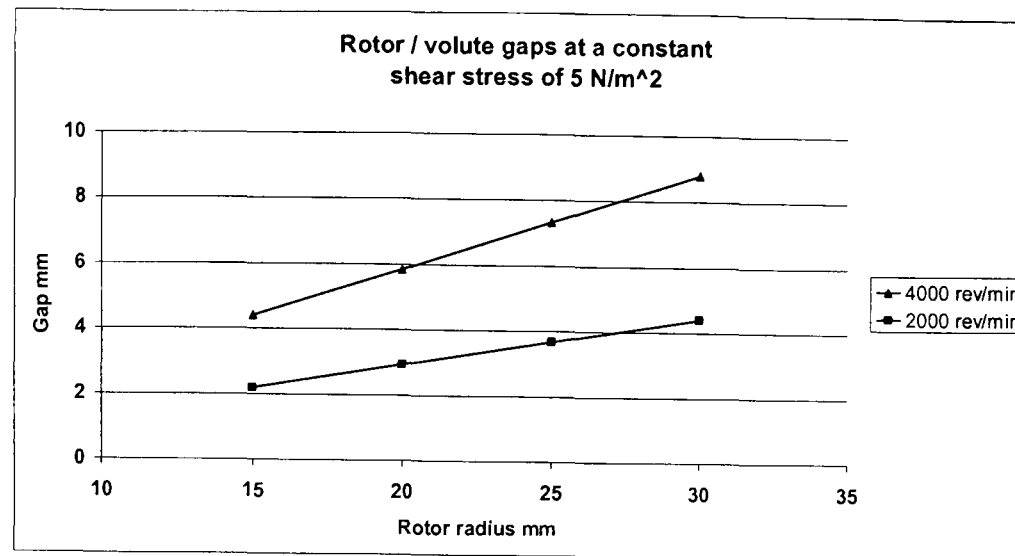


Figure 6.31. Calculated rotor / volute gaps for a constant shear stress of 5 N/m².

6.4.2 Outer casing.

The outer casing of the proposed pump will carry two bearing points supported on struts across the double entry ports.

The casing will be made in two halves fused together such that there is no leakage between them. The inside surface is to be smooth.

The pump will have a double entry system with blood flowing past the bearings. This will avoid any 'dead' areas for blood to stagnate or pool.

6.4.3 Bearings.

Double bearings will be preferred to support the discs. This will prevent any 'whirl' possible when using an overhung bearing system. The bearings will be blood lubricated which follows the method used with the Jarvik 2000 pump. It gives a long lasting bearing system requiring minimum lubrication.

6.4.4 Power and electronics.

It is proposed that the pump will have no electronics 'built in'. This will make the pump itself a 'throw away' item. The pump will fit into an electronics 'cradle' containing the electronics that will generate a rotating magnetic field. The steel disc(s) inside the rotor, attracted by the magnetic field, will rotate with it. The viscous attraction of the blood to the disc(s) will pump the blood following Tesla's principles. Figure 6.32 shows the intended layout.

The pump is intended to be the disposable part, not the electronics. It is assumed that the electronics will be the expensive part of the pumping system and the disc(s) and casing will be the cheap part. This should provide for a cheaper rotary pump than one with electronics 'built in'.

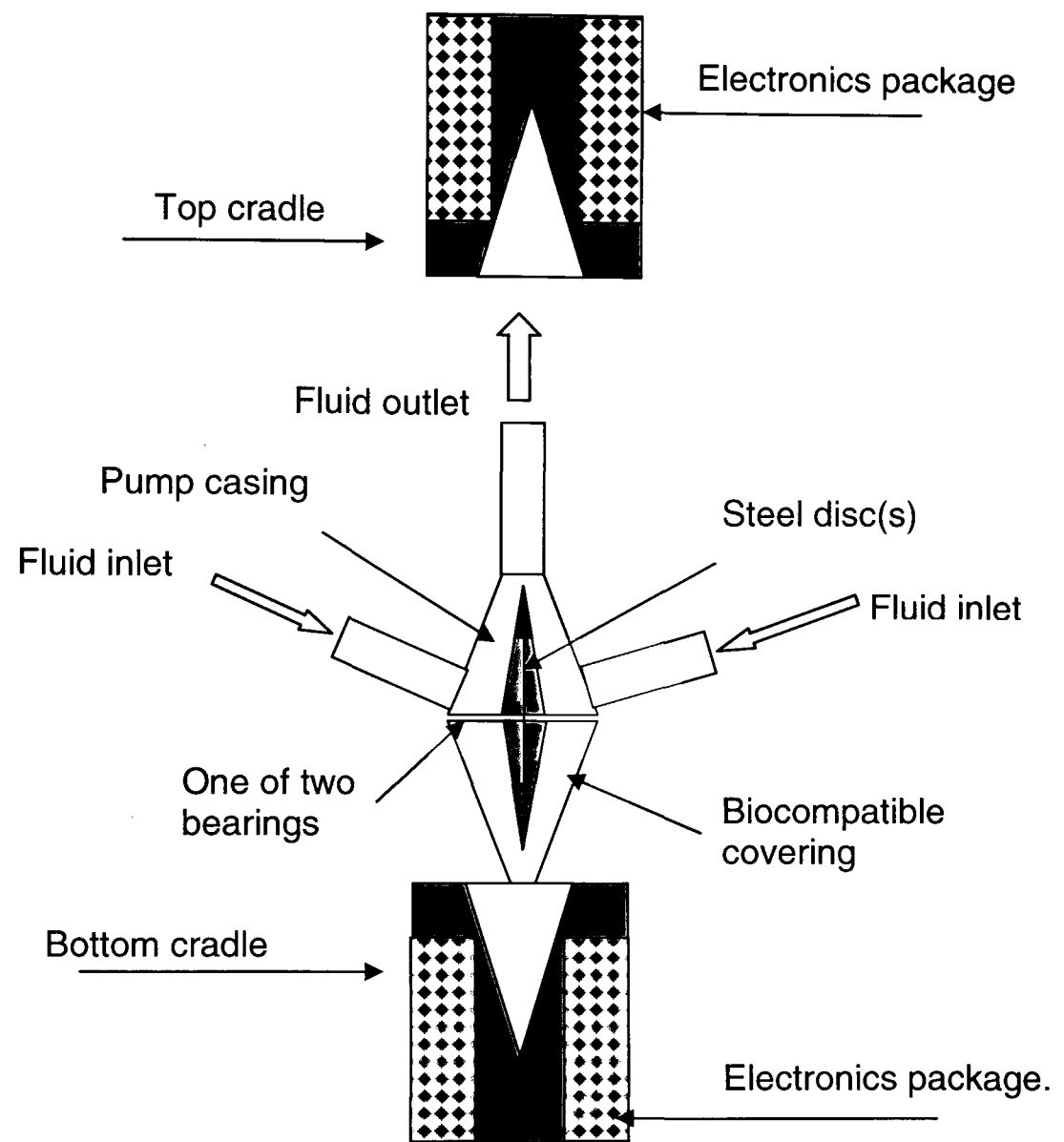


Figure 6.32. Proposed design of new blood pump.

6.5 Flow improvements.

The Tesla pump tested had a vortex swirl pattern at, or close to, at the fluid entry into the pump. Suggestions to stop the vortex are presented as follows:-

- a) Induce a entry pre-swirl in the inlet pipe by fitting angled blades in the inlet pipe before the blood enters the pump.
- b) Allow the blood enter the pump at an angle to the centreline.
- c) Reposition the blood inlet such that blood is already travelling in same linear direction as part of the rotating disc.

Of these possibilities, the repositioning of the entry pipe is considered to be optimum as angled blades are a complication and may lead to blood reactions from any increased stresses.

Figure 6.33 shows a possible configuration where the inlet enters at an angle and is positioned below the centre line of the rotating disc(s).

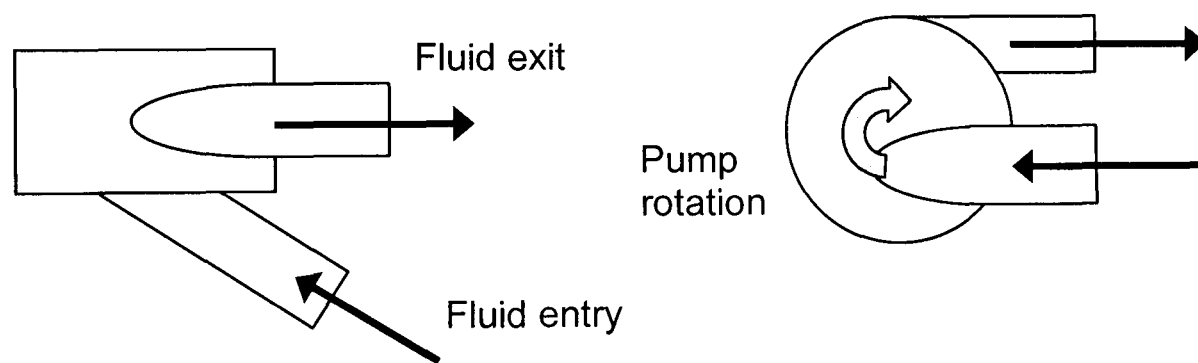


Figure 6.33 Proposed new fluid entry configuration.

The angle and positions of the fluid entry and exit may have to be modified in relation to patient requirements.

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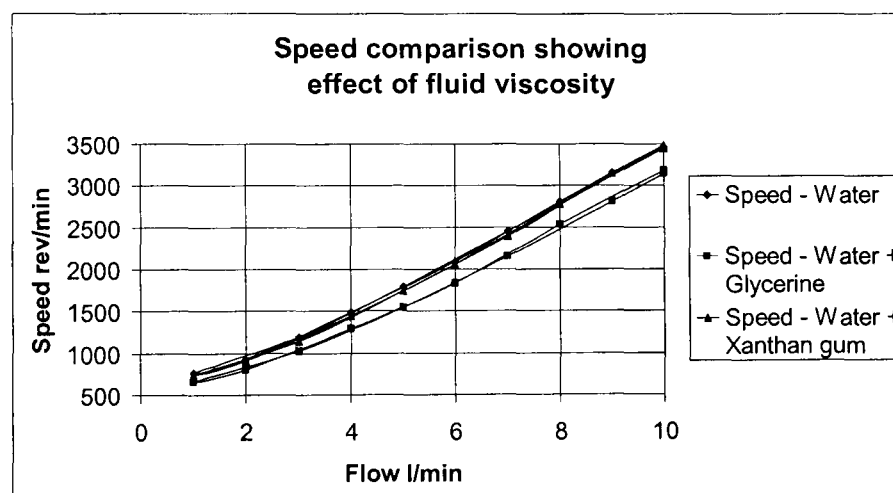
Chapter 7. Conclusions.

This work has shown that a Tesla pump will pump fluids.

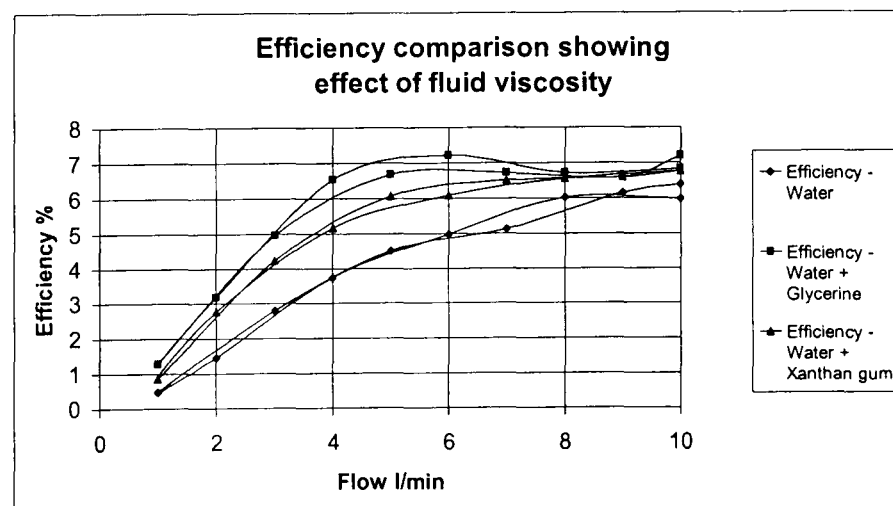
The pump output can be varied depending on the rotor construction and the fluid used. Rotors of different constructions gave different results for the same target performances with the same fluid. The Tesla pump was capable of delivering 10 l/min at 200 mm Hg at 3500 rev/min using water as a test fluid.

The Tesla pump can exceed the delivery and generate pressures greater than the human heart.

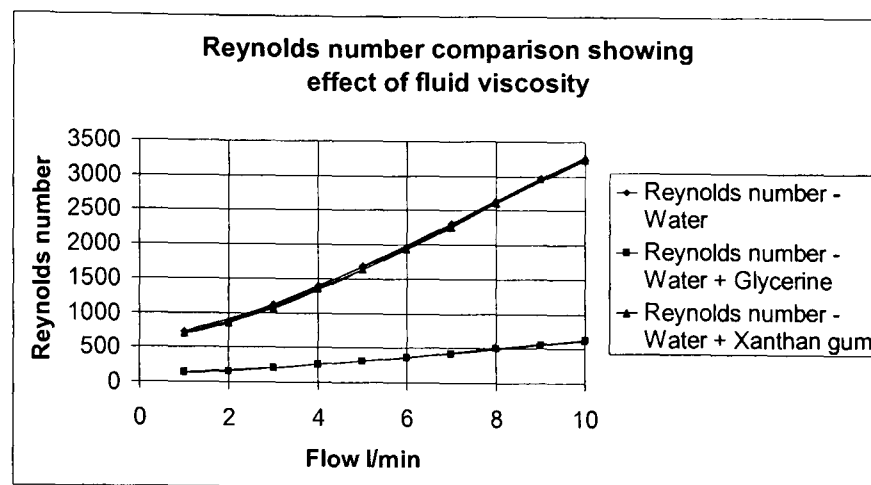
With the same rotor construction the pump performance improves compared to water when using a fluid with a viscosity greater than water. Appendices 14.18, 14.19 and 14.22 are replotted below with added emphasis to show the performance differences with the change in fluid viscosity.



Replot of Appendix 14.18 showing effects of viscosity on pump speed



Replot of Appendix 14.19 showing effects of viscosity on pump efficiency



Replot of Appendix 14.22 showing effect of viscosity on the Reynolds numbers.

The speed / flow plot shows that the speed is lower for the same delivery with the highest viscosity mixture, water – glycerine, compared to water and water – xanthan gum.

The efficiency / flow plot shows that the efficiency tends to be higher with a higher viscosity fluid over most of the flow range tested compared to the other fluids.

The Reynolds number with the water - glycerine mixture is less than one third of the water and water / xanthan gum results for the same delivery. From these figures and assuming a 'limiting' Reynolds number of 2000, the output of the pump can exceed 10 l/min while with the other two fluids the delivery is limited to about 6 l/min.

Although it was not possible to show any results, it is assumed that the pump performance will be worse using a fluid with a viscosity less than water.

The results indicate a better performance with a 'large' gap and 'high' viscosity fluid compared to a 'small' gap and 'low' viscosity fluid although there must be a limit on the size of the gap as shown from the results of rotor 0, when comparing the Reynolds numbers.

The gap may need to be 'matched' to the viscosity to give an optimum result as shown by the results from rotor 0.

The overall efficiency of the Tesla pump is low. The calculated efficiency from Crawford and Rice, (1974) is up to 99% but the highest figure

obtained with the Tesla pump was approximately 10%. Low efficiency is a characteristic of this type of pump.

Disc design affects efficiency. From the calculated results of Crawford and Rice, (1974), discs with an inner / outer diameter ratio of 5 gave a higher efficiency than discs with a diameters ratio of 2. Comparing the shape of the efficiency curve of the Tesla pump tested here to the efficiency curve of a typical rotary pump gave the impression that the Tesla pump was not working at the maximum possible potential. The shape of the efficiency curve relative to speed indicated a higher speed may have given a higher efficiency. The Tesla pump tested here may have been too large for the application and a smaller pump running faster might have been more suitable.

Attempts to promote cavitation were not successful. There was no sign of cavitation during any of the tests although differential pressures of 200 mm Hg were achieved.

Without CFD calculations, it is assumed that the gap between the end discs of the rotor and the wall of the volute will give the maximum fluid shear stress in the pump. With a rotor-to-wall gap of 1.5 mm and a pump speed of 2000 rev/min a shear stress of 15 N/m^2 was obtained. This is greater than the maximum physiological level of stress for blood of 6 N/m^2 . Increasing the gap between the rotor and volute walls would produce a lower shear stress level.

An alternative disc design is proposed that is expected to reduce the stresses to close to the maximum physiological levels for the same pump speed. The effects of this change on pump performance are not known.

The Tesla pump maintained constant pressures at constant speeds up to the maximum flow rate of 10 l/min. This effect is not fully understood as it is different to other pumps where the pressure falls off at constant speed as the flow rate increases. If as suspected, the pump was too large for this application and the delivery had been allowed to increase beyond 10 litres/min, the characteristic of falling pressure with increased delivery may have been obtained.

The pressure produced appeared to be dependant on the speed of the pump. If the assertion by Wiseman, (1994), is correct, a smaller pump running at a higher rotational speed but with the same peripheral speed should give the same pressure. Following this assertion, it should be possible to build a

pump with any disc diameter and rotational speed combination to give the required pressure and by changing the number of discs, to match an output specification. This would make the pump capable of matching an output specification anywhere within its maximum range of pressures and speed.

The Tesla pump produced non dimensional parameter figures that were different to the calculated figures published by Crawford and Rice, (1974). Examples are shown in the table below. It is not understood how a pressure change parameter can be negative for a pump.

<u>Coefficient</u>	<u>Crawford and Rice</u>	<u>Tesla pump results</u>
Efficiency %	0 to 99	0.5 to 12
Re number	0 to 10	100 to 3500
Volume flow rate parameter U_0	0 to 2	0.02 to 0.25
Torque parameter T_1	0 to 1250	2 to 35
Pressure change parameter P_t	-437 to 3.5	0.4 to 1.5

Improvements to the pump design and construction are expected to give practical results that are closer to the calculated results.

From the video and photographs, the fluid flow patterns were not optimum. The generation of a vortex above the fluid inlet and the disturbed flow pattern in the fluid outlet indicates that more work needs to be done on the hydraulic design of the pump. It should be noted that these patterns were only seen after the injection of air into the pump inlet and not during any of the flow tests.

Considering the difficulties with the flow patterns, an angled fluid entry to the pump may be beneficial in order to give a 'pre-swirl' effect to the fluid. Some design outlines are given in chapter 6.

Further practical work needs to be done to explore the potential of this design of pump. The calculated data by Crawford and Rice, (1974), should be a good starting point.

For a pump to be used as a heart-assisting device that is implanted into the patient, the size must be reduced in order that the pump will fit inside the patient. If pressure is controlled by rotor peripheral speed, a smaller pump

will have to run faster in order to maintain a satisfactory performance and careful design will be necessary to ensure acceptable blood stresses.

It is proposed that this type of pump will be a 'single use' pump.

For a pump used outside the body, it is proposed that the pump is built with no motor and no electronics attached to it. If the rotating parts could be attracted by a rotating magnetic field, external to the pump, the pumping action could be achieved without the driving mechanism in contact with blood.

In the literature search there were no details of an automatic control to regulate the blood flow from a heart assisting device.

There does not appear to be a test specification for a blood pump. Work has started on the development of an International Standard but has not yet been completed.

Chapter 8. Future work.

The fluid flow patterns at present can be improved and need to be investigated to prevent flow separation and vortex generation. The use of a Computational Fluid Dynamics, (CFD), package is recommended in order to investigate solutions and provide an estimate of the fluid stress levels. The last point is very important as unacceptable fluid stresses and flow patterns may damage the blood and cause severe problems to the patient.

Parts of the pump to be investigated are:-

a) The cutwater. The reason for the flow 'diving' under the cutwater and the flow separation needs to be investigated and a more acceptable design proposed.

b) The vortex generated in the output flow appears to be related to the cutwater or the volute wall. The cause of the vortex needs to be established and a solution provided to eliminate this problem.

c) The efficiency of the pump needs to be improved. The work by Crawford and Rice, (1974), indicated that the calculated efficiency reached a maximum with a disc diameter ratio of about 5, (this is the ratio of the diameter of the inner hole to the outer diameter of the disc). Using this work as a starting point, the design of the discs should be investigated to confirm this finding and suggest improvements.

d) The CFD work should be used to investigate the flow between the discs and around the cutwater and to calculate the fluid stresses. Other design changes need to be investigated as necessary.

e) Investigations are needed to determine the potential of the pump to heat the fluid it is pumping. Excessive temperatures can damage blood and changes must be made to ensure that the pump does not increase the blood temperature above acceptable limits.

f) Any design changes must provide acceptable fluid stress levels. Previous work indicates that stress levels of approximately 5 N/m² would be acceptable.

APPENDIX 1. THE ELEMENTS OF BLOOD.

<u>Characteristics</u>	<u>Red cells</u>	<u>White cells, total white cells, 5 – 9 thousand per mm³</u>					<u>Platelets</u>
Number	(4.5 – 5) million per mm ³	<u>Neutrophils.</u> 65–75% of all white cells	<u>Eosinophils.</u> 2-4% of all white cells	<u>Basophils.</u> 1% of all white cells	<u>Lymphocytes.</u> 20-25% of all white cells	<u>Monocytes.</u> 2-6% of all white cells	250 – 300 thousand per mm ³
Average size, (diameter in microns)	7 - 8	10 - 12	10 - 14	9 - 12	6 - 12	12 - 15	2 - 3
Appearance of nucleus	None	Dark, up to 5 lobes	Pale, 2-3 lobes	Pale, 2-3 lobes	Very dark, slightly indented	Light, indented	None
Appearance of cytoplasm	Red	Clear, many tiny purple granules	Pale, moderate number of red granules	Pale, a few coarse dark blue granules	Blue, agranular, scant	Blue-grey agranular	Very dark
Source	Bone marrow	Bone marrow	Bone marrow	Bone marrow	Lymphoid tissue	Lymphoid tissue	Bone marrow, (megakaryocyte)
Motility	None	Marked	Little	Marked	Marked	Marked	None
Phagocytic ability	None	Marked	Negligible	Negligible	Negligible	Marked	None
General function	O ₂ and CO ₂ transport	Phagocytosis	??? allergy?	??? anticoagulant	Protective, immunities, antibody carrier	Phagocytosis	Blood clotting
Miscellaneous properties	Shape of bi concave disc				Found in connective tissue in great quantities	Found in connective tissue in great quantities	Cell fragment

After:- Essential Human Anatomy and Physiology, B.R. Landau, Scott, Foresman &Co., Glenview, Illinois, USA, 1976, ISBN 0-673-05989-8

APPENDIX 2. PHYSIOLOGICAL FLOWN PARAMETERS IN THE HUMAN CIRCULATION.

Vessel	Diameter $\text{m} \cdot 10^{-3}$	Volume flow rate $\text{m}^3/\text{s} \cdot 10^{-3}$			Mean linear velocity $\text{m/s} \cdot 10^{-3}$			Reynolds number	Wall shear rate l/s			Wall shear stress N/m^2		
		min	max	mean	min	max	mean		min	max	mean	min	max	mean
Ascending aorta	23.0-43.5	-	-	364	-	-	245-876	3210-6075	-	-	45-305	-	-	0.16-1.07
Femoral artery	5	-6.9	23.1	3.7	-350	1175	188	283	-560	1885	302	1.96	6.59	1.06
Common carotid	5.9	2.7	10.6	5.1	99	388	187	332	134	526	253	0.47	1.84	0.89
Carotid sinus	5.2	1.8	6.9	3.3	85	325	156	244	130	500	240	0.46	1.69	0.84
External carotid	3.8	0.9	3.7	1.8	83	327	157	180	175	687	331	0.61	2.41	1.16
Small arteries	0.3	-	-	$3.5 \cdot 10^{-3}$	-	-	50	2.3	-	-	1335	-	-	4.67
Arterioles	0.025	-	-	$2.5 \cdot 10^{-6}$	-	-	5	$3.8 \cdot 10^{-2}$	-	-	1600	-	-	5.6
Capillaries	0.006	-	-	$7.5 \cdot 10^{-9}$ to $5.7 \cdot 10^{-8}$	-	-	0.2-1.5	$3.6 \cdot 10^{-4}$ to $2.7 \cdot 10^{-3}$	-	-	265-2000	-	-	-
Thoracic inferior vena cave	20	-	-	34-50	-	-	107-160	323-482	-	-	43-64	-	-	0.15-0.22

Goldsmith H.L., Turitto V.T., (1986)

APPENDIX 3. INVENTORY OF BLOOD PUMPS.

<u>Displacement pumps</u> <u>Pulsatile VAD</u>		<u>Rotary blood pumps. Continuous flow devices</u>						<u>Total artificial heart</u> <u>Heart replacement</u>	
		<u>Axial pumps</u>		<u>Radial pumps</u>		<u>Diagonal pumps</u>			
*HeartMate IP/VE	Novacor	Hemopump	*Jarvik 2000	Biopump	Capiox	*VentrAssist	HeartQuest	*Abiomed TAH	Akutsu III TAH
LionHeat	Thoratec (Pierce-Donachy)	Sun Waseda IVAP	DeBakey	Delphin	Isoflow (Lifestream)	DeltaStream	HIA Microdiagonal	Cardiowest, TAH (Jarvik 7)	Liotta TAH
Medos / HIA	*Berlin Heart	*HeartMate II (Axipump)	Impella	Rotodynamic pump	Gyro Pump			PennState TAH	Utah TAH (EHTAH)
*Abiomed BVS 5000	Nippon-Zeon VAD	Valvo Pump	Streamliner	RotaFlow	HiFlow			Phildelphia Heart	Nimbus TAH (E4T)
Toybo VAD	ALVAD (Model 7)			Nikkiso Pump	AB 180			Baylor TAH	*Undulation Pump
BCM	Cora Pump			*Vienna centrifugal	Evaheart				
*Roller Pump	*Intra-Aortic balloon pump			Abiomed CF	Kriton Pump				
0HeartSaver	PUCA Pump			MSCP	*HeartMate III				
*Abiocor									

The pumps denoted by * are described in this report.

Blood pumps for circulatory support
Perfusion 2000; 15: 295-311.

APPENDIX 4.

ARTIFICIAL HEART TYPES AND APPLICATIONS.

Perfusion 2000.

The pumps denoted by * are described in this report.

Name	Manufacturer	Type	Flow	Capacity	Speed	Duration	Power	Application	Usage
*Roller (Peristaltic)	Watson-Marlow- Bredel	Pulse	Pulsatile	Not stated	Not applicable	4 hours	Electric motor	Adult	Hospital
*IntraAortic	Datascope Medical	Pulse	Pulsatile	Not applicable	Not applicable	Short term	Pneumatic	Adult	Hospital
*Abiomed BVS 5000	Abiomed BV, Europe	Pulse	Pulsatile	100 ml vol.	Not applicable	Short term	Pneumatic	Adult	Hospital
*HeartMate	Thermo Cardiosystems	Pulse	Pulsatile	2-10 l/min	Demand driven	90 days	Pneumatic/elec.	Adult	Hospital / public
*Berlin	Mediport Kardiotechnik	Pulse	Pulsatile	10-50 ml. vol.	Not applicable	1 year +	Pneumatic	Infant-Adult	Hospital / public
*Abiocor	Abiomed	Pulse	Pulsatile	Not stated	Not applicable	Short term	Batteries	Adult	Hospital / public
*Atom pump	Never made	Pulse	Pulsatile	Not stated	Not applicable	30+ years	Plutonium 238	Adult	Never used
*Jarvik 2000	Jarvik Heart Inc.	Rotary	Axial	5-6 l/min	Up to 12,000 rev/min	Long term	Batteries	Adult	Hospital / public
*AB 180	Abiomed	Rotary	Centrifugal	6 l/min	2500-4500 rev/min	14 days	12V motor	Adult	Hospital
DeBakey	Baylor College/NASA	Rotary	Axial	5 l/min	10,000 rev/min	Not stated	Batteries	Adult	Hospital
Terumo	Terumo	Rotary	Radial	5-6 l/min	2,000 to 3,000 rev/min	48-72 hours	Batteries	Adult	Hospital
Thoratec	Thoratec Europe	Pulse	Pulsatile	65 ml vol.	Demand driven	38-80 days	Pneumatic	Adult	Hospital
Novacor	Novacor Baxter	Pulse	Pulsatile	8.5 l/min	Demand driven	9 months	Batteries	Adult	Hospital / public
Medos	Medos Medizintechnik	Pulse	Pulsatile	0.5-9.5 l/min	Fixed and demand	2 weeks +	Pneumatic/elec.	Infant-Adult	Hospital

Appendix 5.

Mechanical Properties of Stainless Steel Surgical Implants

Composition of 316 and 316L steels. ^a(by weight, %.)

<u>ELEMENT.</u>	<u>GRADE 1.</u>	<u>GRADE 2.</u>
Carbon	0.08 max	0.03 max
Manganese	2.00 max	2.00 max
Phosphorus ^b	0.03 max	0.03 max
Sulphur	0.03 max	0.03 max
Silicon	0.75 max	0.75 max
Chromium	17.00 – 20.0	17.00 – 20.00
Nickel	12.00 – 14.00	12.00 – 14.00
Molybdenum	2.00 – 4.00	2.00 – 4.00

^a Ref :- Annual book of ASTM Standards, Part 46, American Society for Testing and Materials, Philadelphia, 1980, p578.

^b Slight variations are given, (0.025 max.), for special quality steels, (F318 and F319 of ASTM).

Appendix 5. (continued)

Mechanical Properties of Stainless Steel Surgical Implants.^a

	ULTIMATE TENSILE STRENGTH (MIN.)	YIELD STRENGTH, (0.2% OFFSET). (MIN.)	ELONGATION, 2 INCH, (50.8 MM).	ROCKWELL HARDNESS.
<u>Condition</u>	PSI, (Mpa)	PSI, (Mpa)	Min. (%)	Max.
<u>Grade 1</u> <u>(type 316)</u>				
Annealed	75,000, (515).	30,000, (205)	40	95 HRB
Cold finished	90,000, (620)	45,000, (310)	35	----
Cold worked	125,000, (860)	100,000, (690)	12	300 - 350
<u>Grade 2</u> <u>(type 316L)</u>				
Annealed.	73,000, (505)	28,000, (195)	40	95 HRB
Cold finished	88,000, (605)	43,000, (295)	35	---
Cold worked	125,000, (860)	100,000, (690)	12	---

^a Ref: - Annual Book of ASTM Standards, Part 46, American Society for Testing and Materials, Philadelphia, 1980, p579

Appendix 6

Plastics and Applications.

<u>Name of plastic.</u>	<u>Application.</u>
Polyvinylchloride, (PVC)	Blood and solution bags, surgical packaging, IV sets, dialysis devices, catheter bottles, connectors and cannulae.
Polyethylene, (PE)	Pharmaceutical bottle, non-woven fabric, catheter pouch, flexible container and orthopaedic implants.
Polypropylene, (PP)	Disposable syringes, blood oxygenator membrane, suture, non-woven fabric and artificial vascular grafts.
Polymethylmetacrylate, (PMMA)	Blood pump and reservoirs, membrane for blood dialyser, implantable ocular lens and bone cement.
Polystyrene, (PS)	Tissue culture flasks, roller bottles and filter wares.
Polyethylenterephthalate, (PET)	Implantable suture, mesh, artificial vascular grafts and heart valves.
Polytetrafluoroethylene, (PTFE)	Catheter and artificial vascular grafts.
Polyurethane, (PU)	Film tubing and components.
Polyamide, (nylon)	Packaging film, catheters, sutures and mould parts.

Project report K. Cordes,

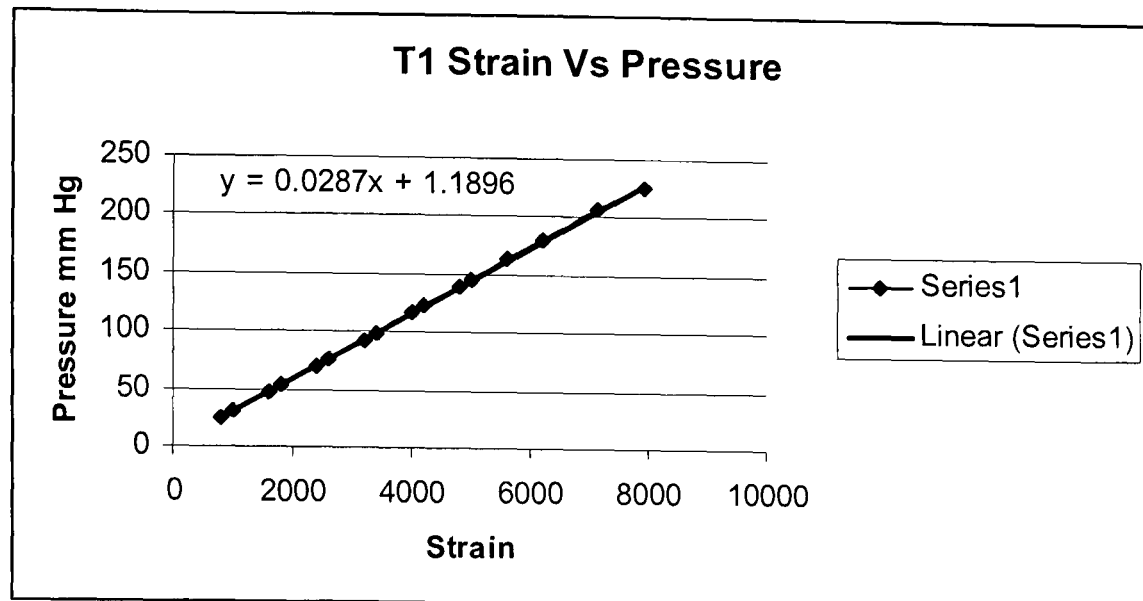
APPENDIX 7.1 CALIBRATION FIGURES FOR THE PRESSURE TRANSDUCERS

Strain indicator, model P-530, ser. No. 005384. Gauge factor = 2.11. Balance = 4.91 1mm Hg*0.491/25.4=0.0193307 lbf/in².

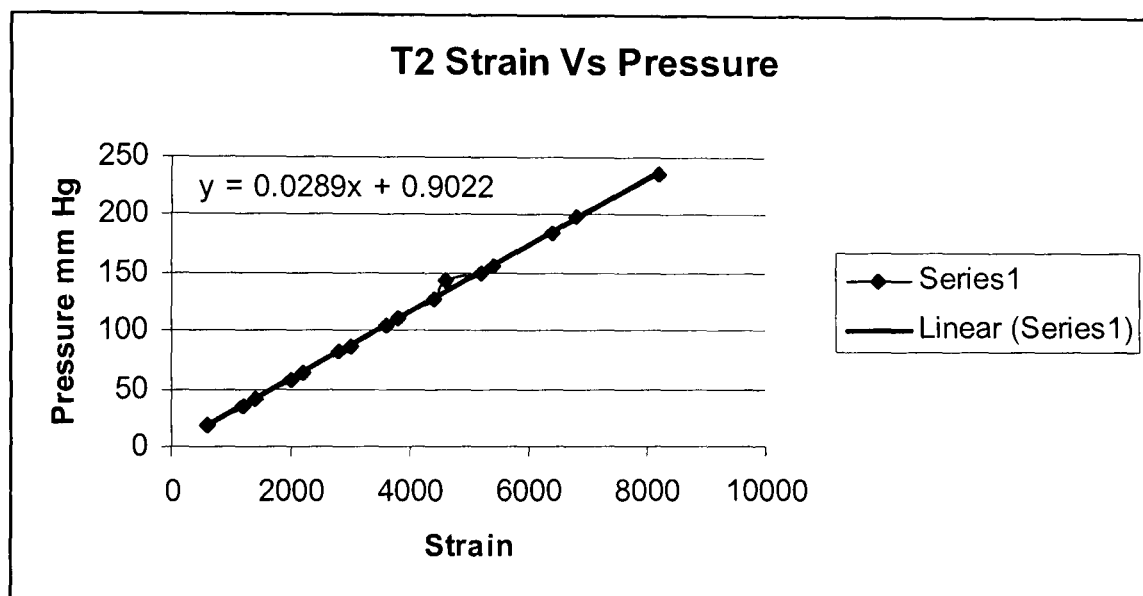
Vishay switch and balance unit. Model SB – 1, Ser. no. 025761

Observed readings Press cm Hg						Strain						Correction factor 0.019331			
												Calculated pressures			
Trans. 1	Trans. 2	Trans. 3	Trans. 1	Trans. 2	Trans. 3	Trans. 1	Trans. 2	Trans. 3	T1 mmHg	T1 lbf/in ²	T2 mmHg	T2 lbf/in ²	T3 mmHg	T3 lbf/in ²	
49.4	26.9	50.0	26.4	52.2	24.2	7900	8200	9400	225	4.349	236	4.562	280	5.41	
48.9	28.2	48.4	28.6	51.4	25.5	7100	6800	8900	207	4.002	198	3.828	259	5.01	
47.6	29.6	47.8	29.3	50.8	26.4	6200	6400	8400	180	3.480	185	3.576	244	4.72	
46.8	30.5	46.4	30.8	50.2	27.0	5600	5400	8000	163	3.151	156	3.016	232	4.48	
46.0	31.5	46.2	31.2	49.8	27.6	5000	5200	7700	145	2.803	150	2.900	222	4.29	
45.7	31.8	45.5	31.1	49.3	28.2	4800	4600	7300	139	2.687	144	2.784	211	4.08	
45.0	32.7	45.2	32.4	48.9	28.7	4200	4400	7000	123	2.378	128	2.474	202	3.90	
44.7	33.1	44.4	33.4	48.6	28.4	4000	3800	6800	116	2.242	110	2.126	202	3.90	
43.8	34.0	44.1	33.7	48.2	29.5	3400	3600	6500	98	1.894	104	2.010	187	3.61	
43.6	34.3	43.3	34.7	47.9	29.8	3200	3000	6300	93	1.798	86	1.662	181	3.50	
42.8	35.3	43.1	35.0	47.5	30.3	2600	2800	6000	75	1.450	81	1.566	172	3.32	
42.6	35.6	42.3	35.9	47.3	30.6	2400	2200	5800	70	1.353	64	1.237	167	3.23	
41.8	36.5	42.0	36.2	47.0	30.9	1800	2000	5600	53	1.025	58	1.121	161	3.11	
41.5	36.8	41.3	37.2	46.6	31.4	1600	1400	5300	47	0.909	41	0.793	152	2.94	
40.8	37.8	41.0	37.5	46.3	31.7	1000	1200	5100	30	0.580	35	0.677	146	2.82	
40.5	38.1	40.3	38.4	46.0	32.0	800	600	4900	24	0.464	19	0.367	140	2.71	
				45.8	32.3	200	400	4700	0	0.000	0	0.000	135	2.61	
				45.5	32.7	100	0	4500	0	0.000	0	0.000	128	2.47	
40.2	38.5	40.2	38.5	45.3	33.0	530	550	4300	17	0.329	17	0.329	123	2.38	
				45.0	33.4			4100					116	2.24	
				44.8	33.6			3900					112	2.17	
				44.6	33.9			3700					107	2.07	
				44.3	34.2			3500					101	1.95	
				44.0	34.5			3300					95	1.84	
				43.7	35.0			3000					87	1.68	
				43.4	35.3			2800					81	1.57	
				43.1	35.5			2600					75	1.45	
				42.8	35.9			2400					69	1.33	
				42.5	36.3			2200					62	1.20	
				42.3	36.6			2000					57	1.10	
				42.0	36.8			1800					52	1.01	
				41.8	37.2			1600					46	0.89	
				41.5	37.5			1400					40	0.77	
				41.3	37.8			1200					35	0.68	
				41.1	38.1			1000					30	0.58	

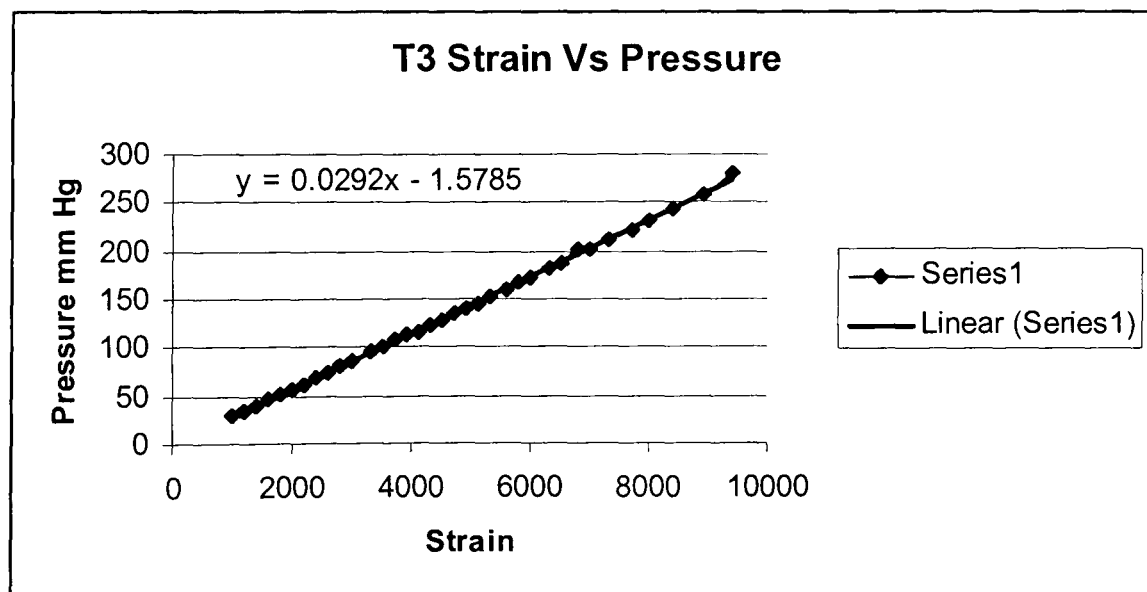
Appendix 7.2. Strain Vs pressure for three transducers.



Transducer 1. Strain Vs pressure.



Transducer 2. Strain Vs pressure.



Transducer 3. Strain Vs pressure.

Appendix 8.

Electric motor and control equipment.

Motor.

Parvalux type no. PM 50, 24vDC, 255 W, 5000 rev/min.

Continuous rating, built to International Organisation for Standardization, (ISO), specification number ISO 9002.

Tacho generator fitted to non-drive end of motor.

Power supply.

Radio Spares - unregulated power supply, RS part no 597-582.

Input 115/230 V ~ 50/60 Hz

Output 24V DC, 288 W.

Control unit.

Parvalux 2020 PWM manual controller,

Input 12/24 V DC, 0-12 Amps.

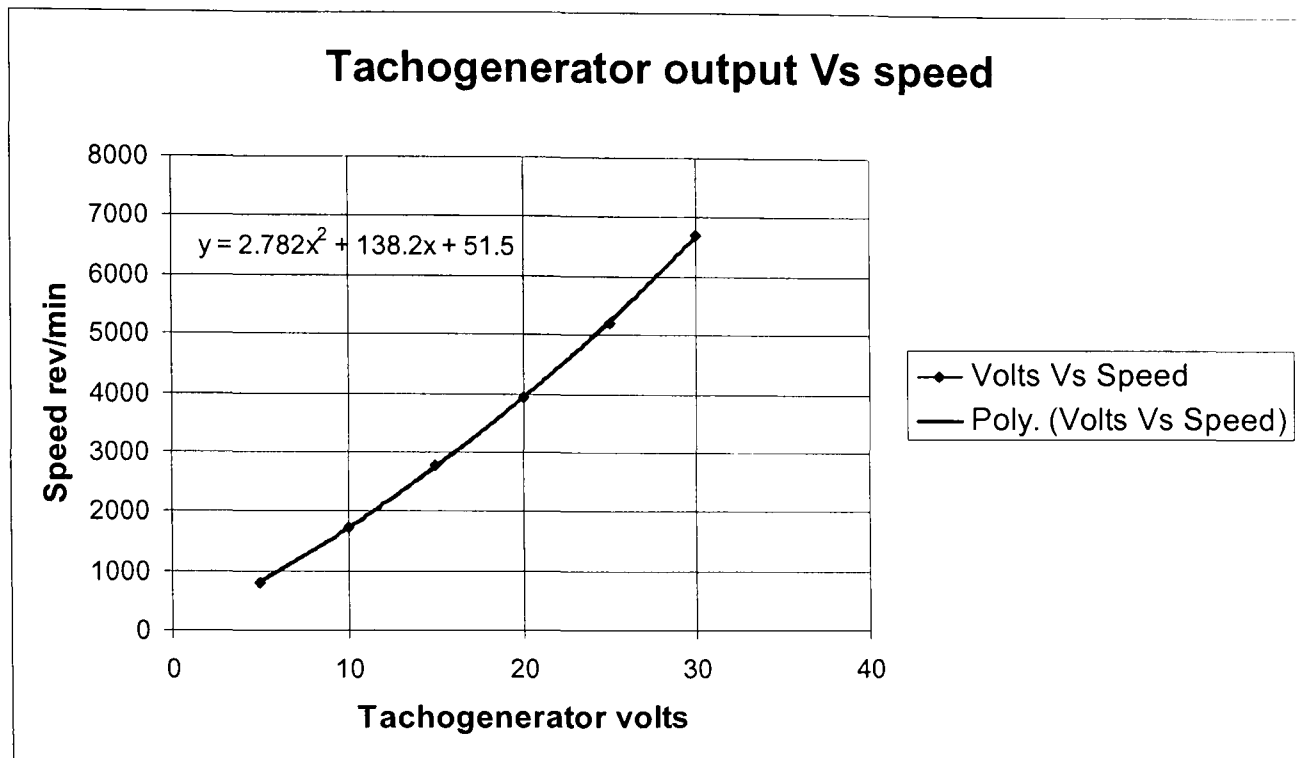
Output 0-24 V DC, 12 Amps. maximum rating.

Contact details.

Electric motor purchased from Parvalux Electric Motors Ltd., Bournemouth, England. [<http://www.parvalux.co.uk>],

Control equipment from RS Components, PO Box 99, Corby, Northants, NN17 9RS. (rswww.com)

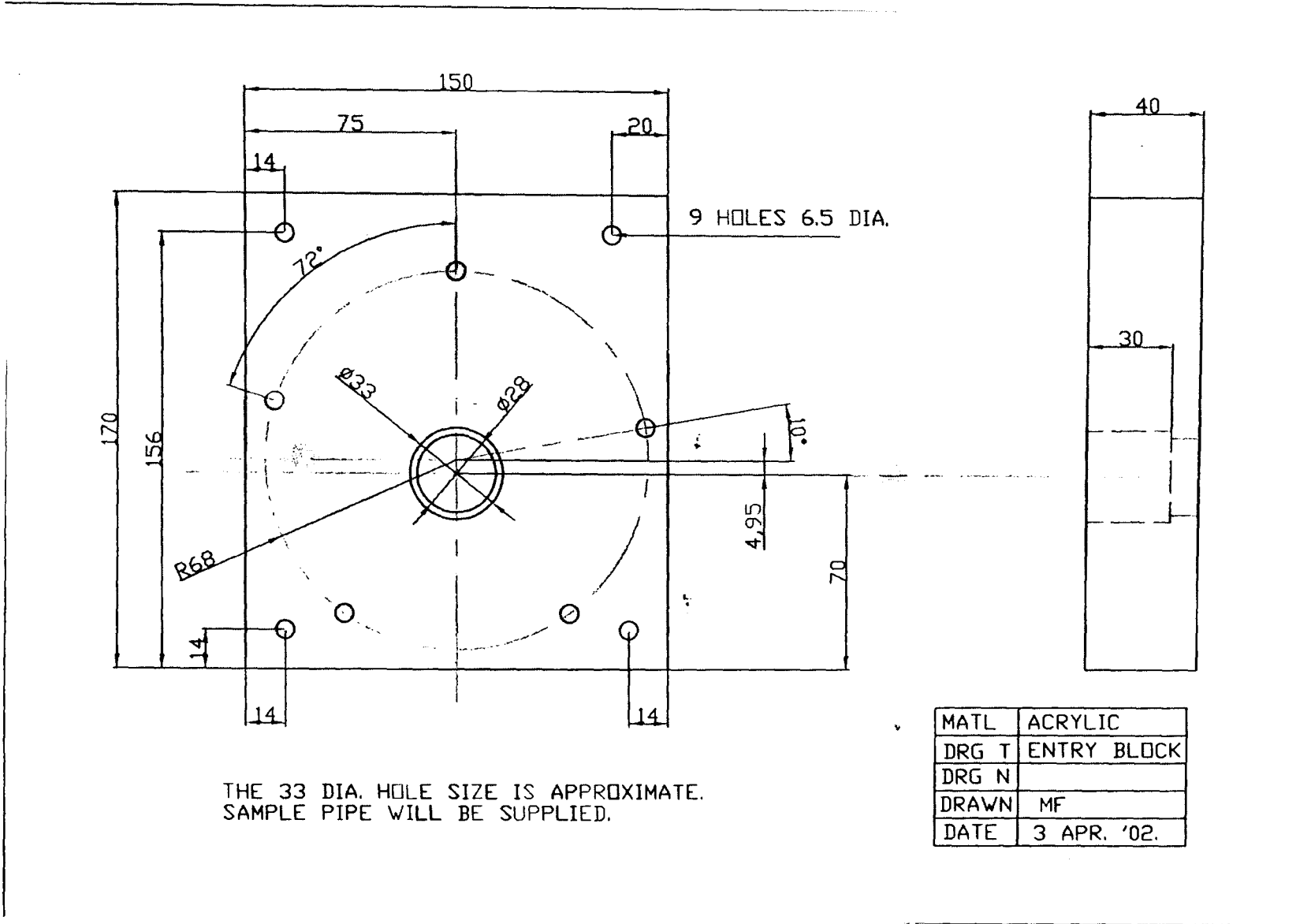
Appendix 9. Calibration of tachogenerator.



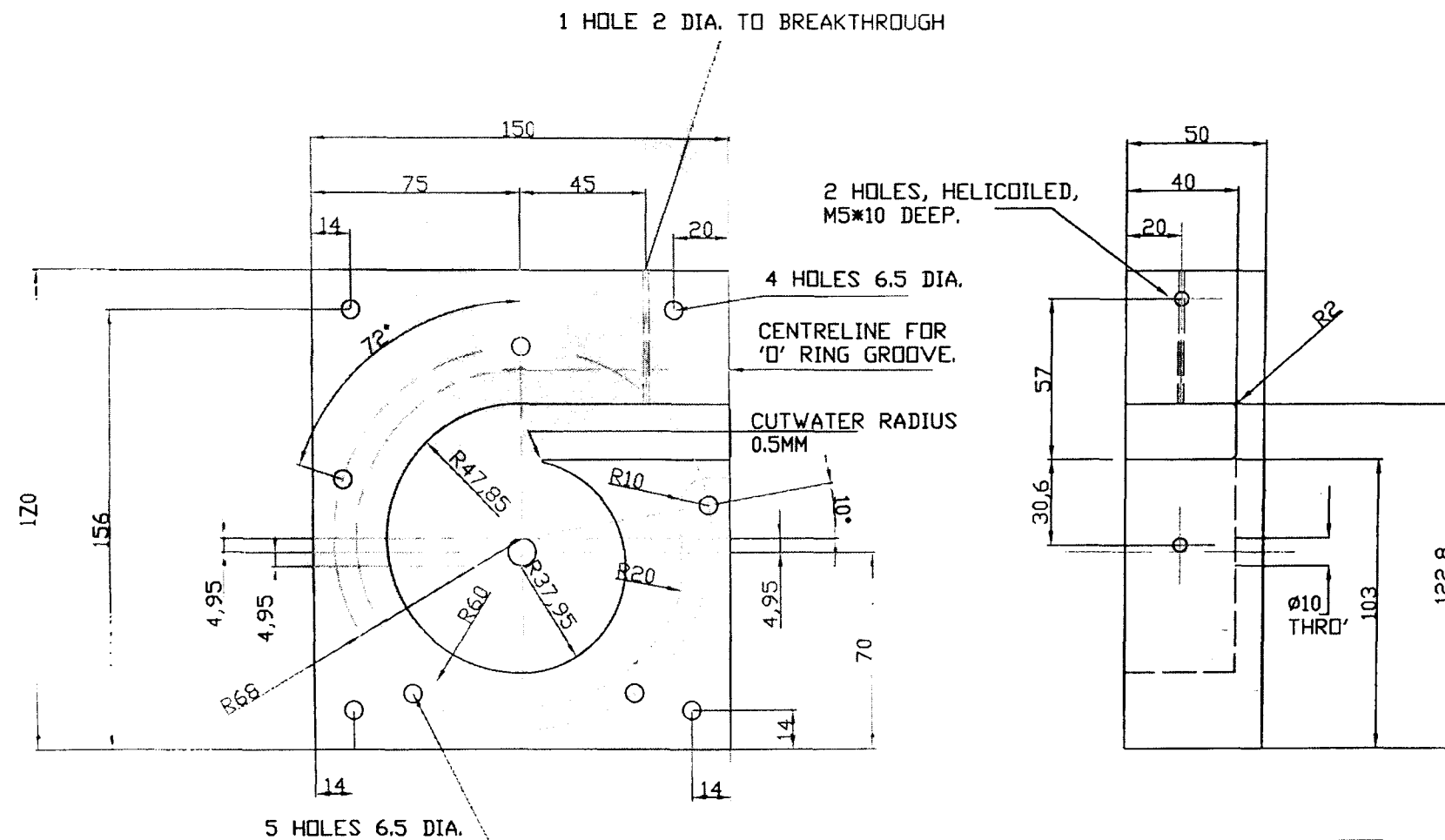
The equation shown below was used in all the calculations of speed from the tachogenerator voltage.

$$\text{Speed (rev / min)} = (2.782 (\text{tachovolts})^2) + (1.382 \text{ tachovolts}) + 51.5$$

Appendix 10.1, Entry block



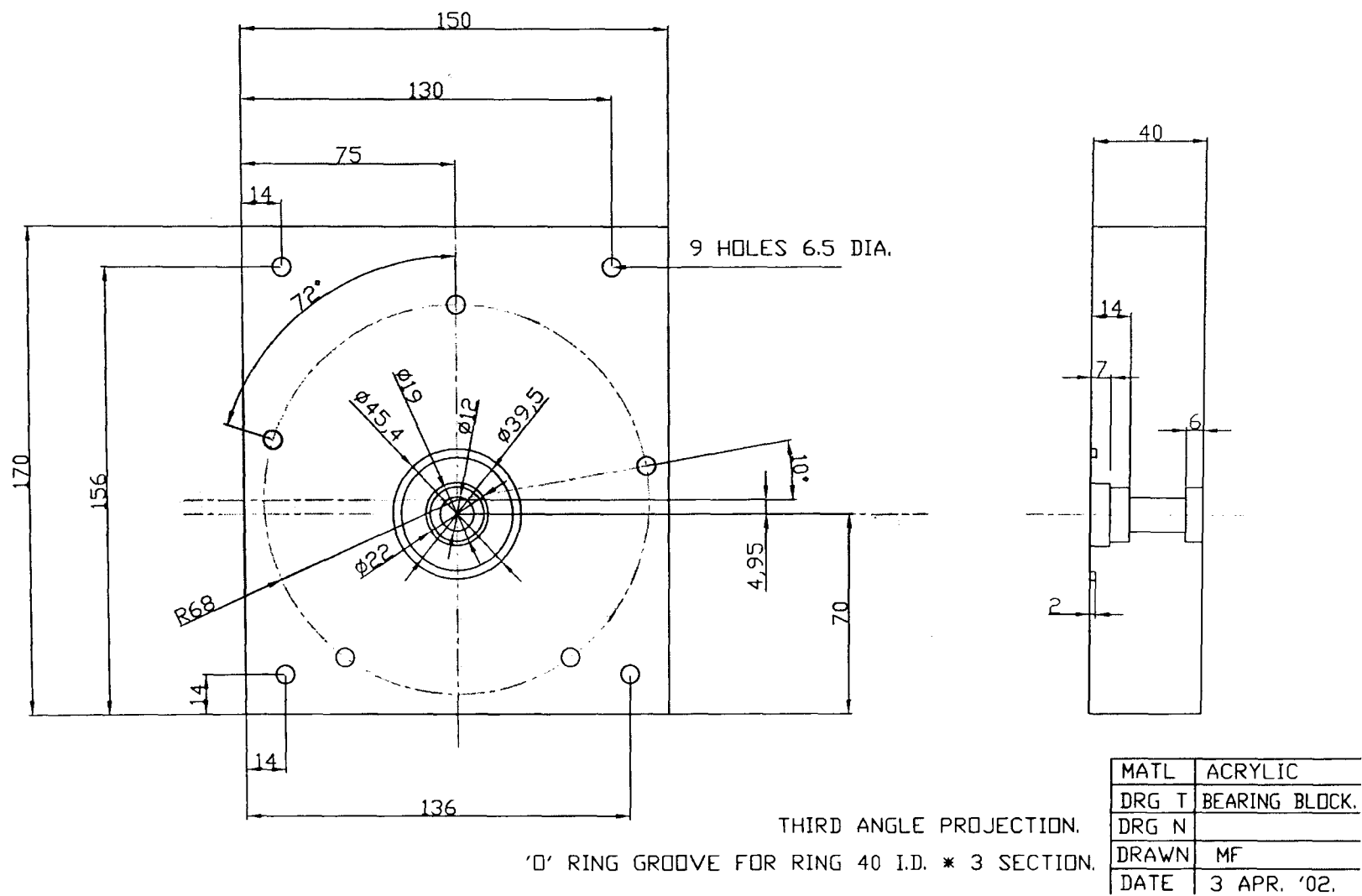
Appendix 10.2, Volute 40.



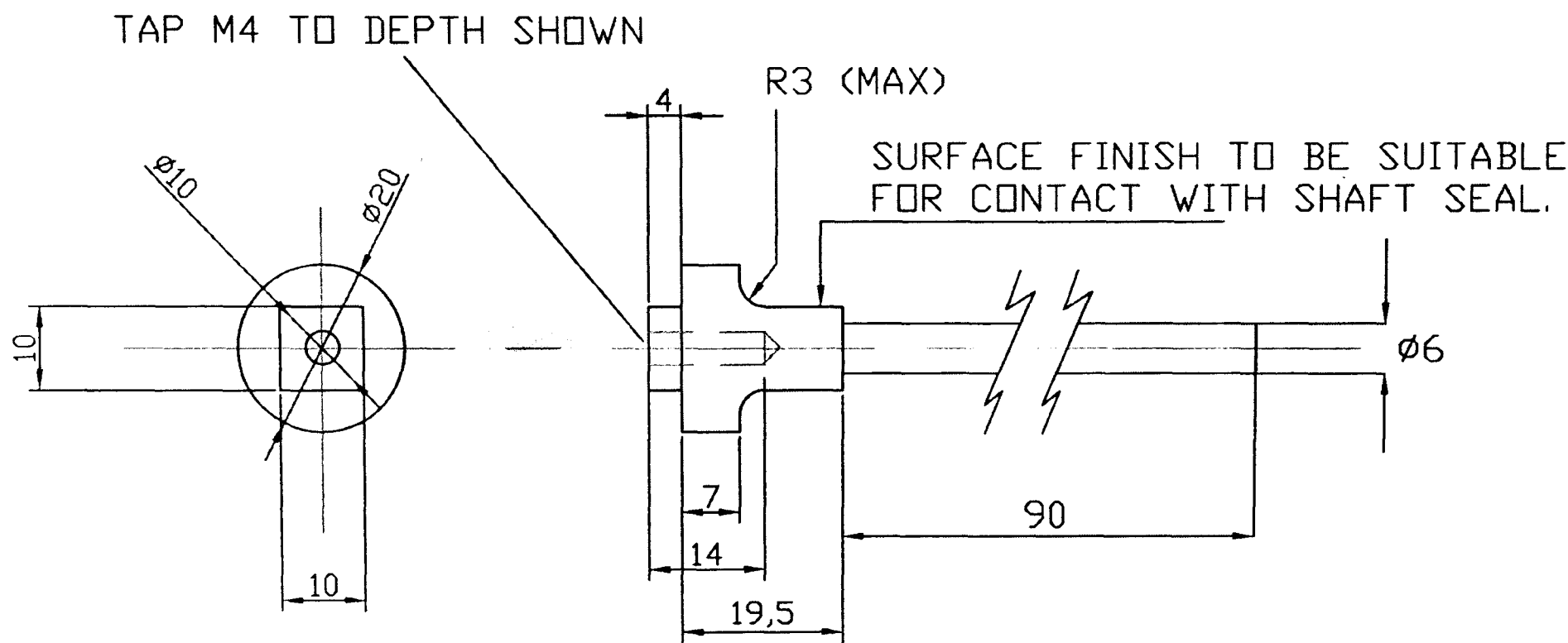
ALL MACHINED SURFACES TO BE POLISHED TO ALLOW THE PASSAGE OF LIGHT 'O' RING GROOVE TO BE SUITABLE FOR 3MM NITRILE RUBER CORD. THE SHAPE OF THE VOLUTE IS MADE UP FROM TWO CIRCULAR ARCS. EACH ARC IS TAKEN FROM THE VERTICAL CENTRELINE OF THE BLOCK.

	ACRYLIC
T	VOLUTE 40
DATE	28 JUNE '02.

Appendix 10.3, Bearing block

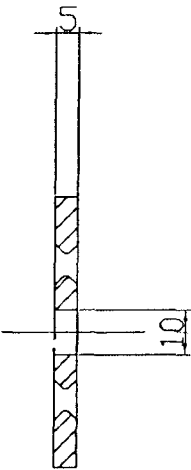
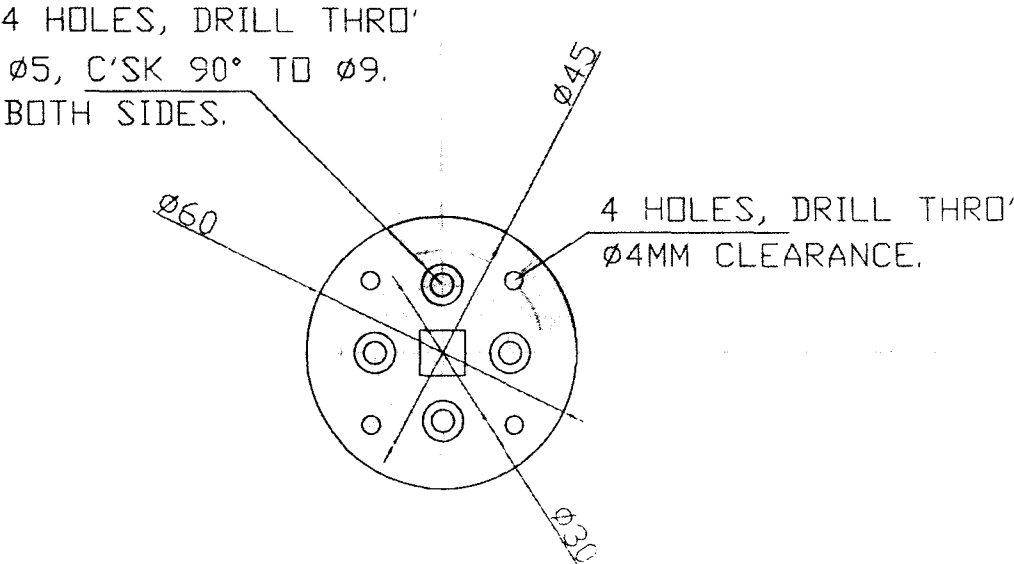


Appendix 10.4, Shaft.



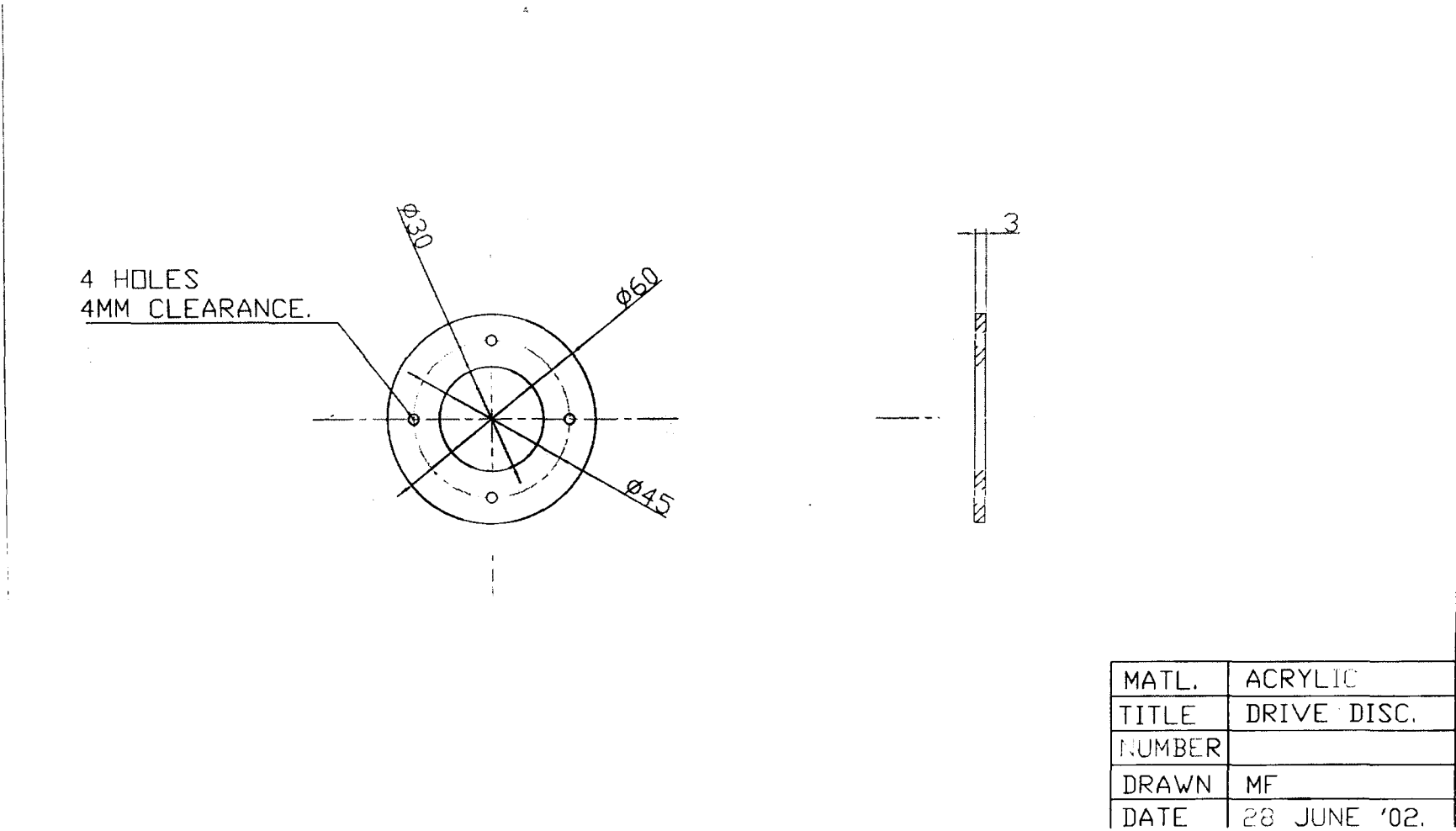
DGR T	SHAFT
DRG N	
DRAWN	MF
DATE	6 DEC '06

Appendix 10.5, Mounting disc.

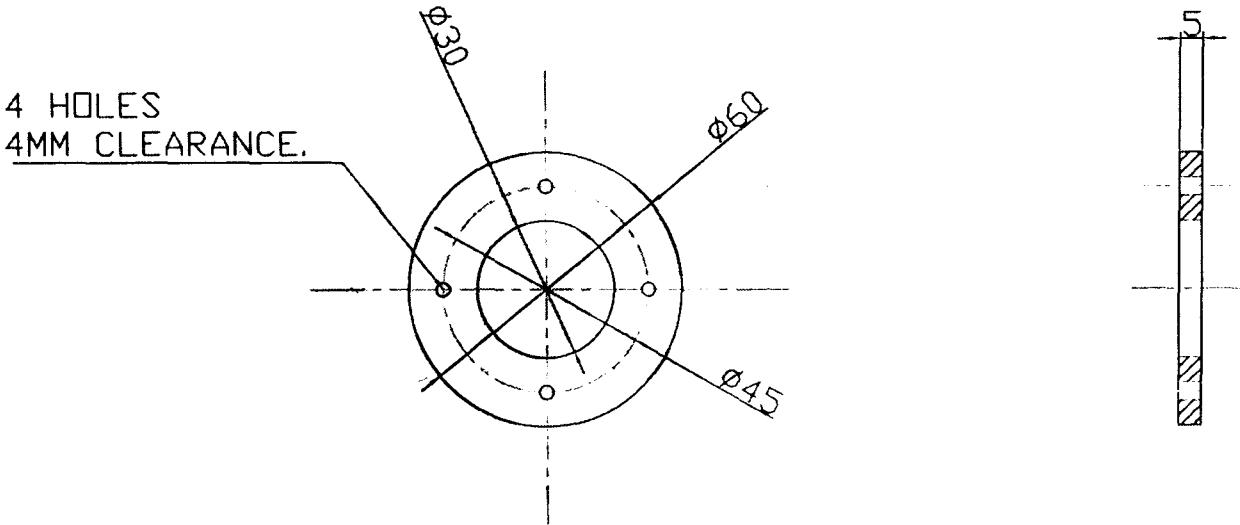


MATL.	ACRYLIC.
DRG T	MOUNTING DISC 2.
DRG N	
DRAWN	MF
DATE	'02.

Appendix 10.6, Drive disc

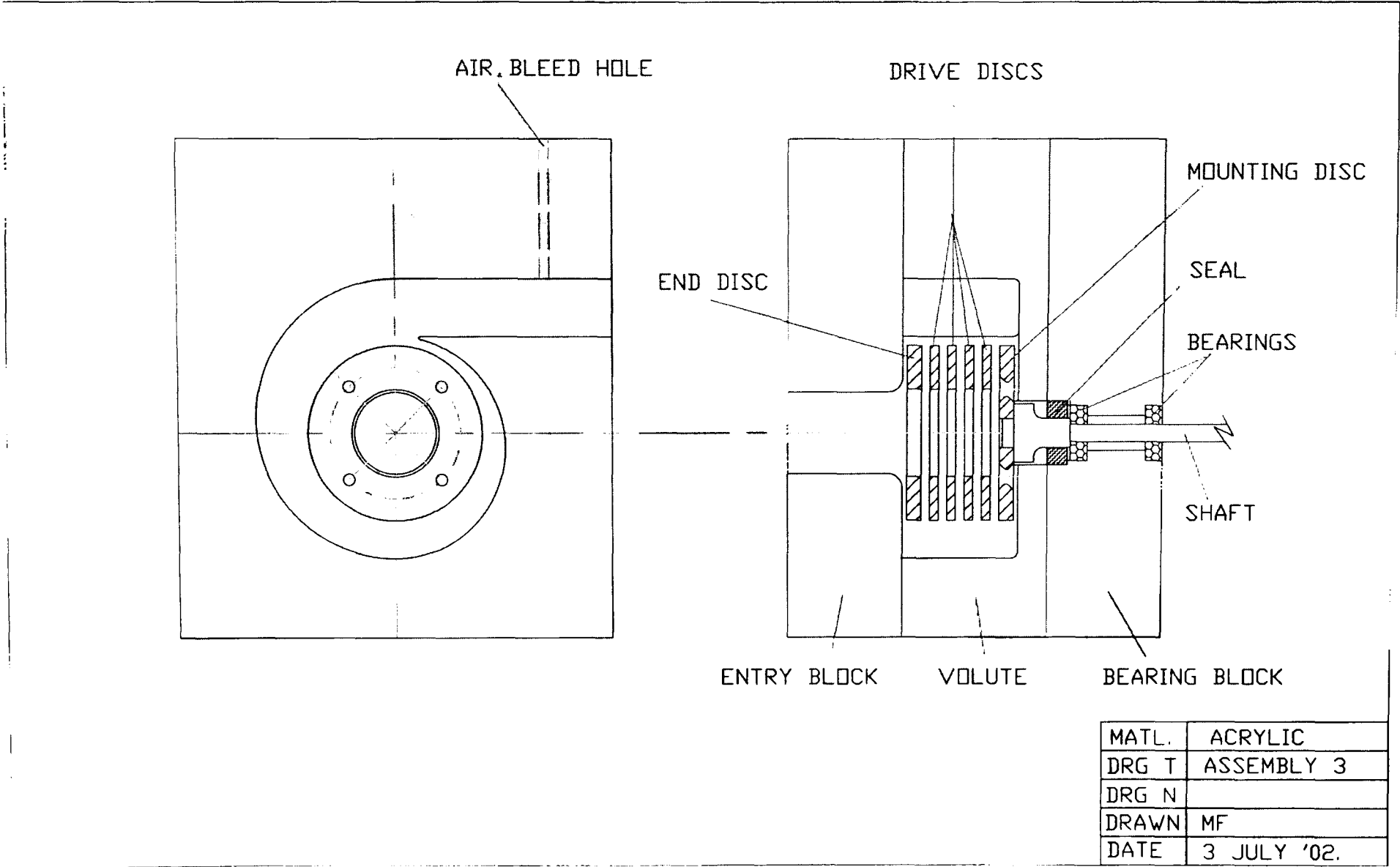


Appendix 10.7, End disc

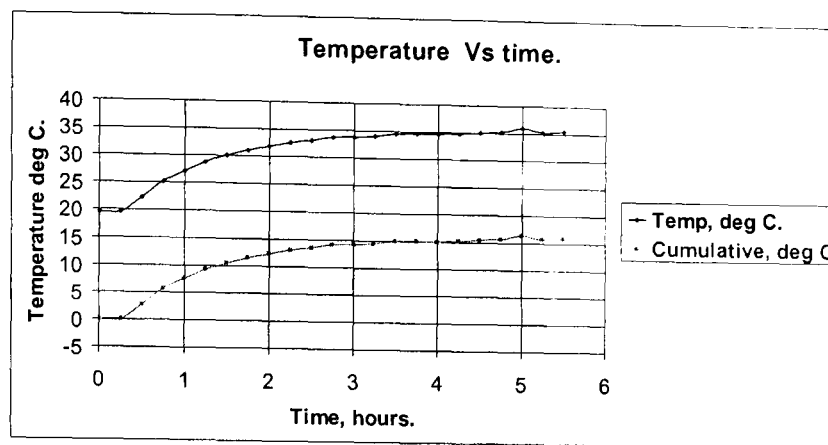


MATL.	ACRYLIC
TITLE	END DISC.
NUMBER	
DRAWN	MF
DATE	6 NOV '02

Appendix 10.8, Assembly.



Appendix 11. Results of temperature test.



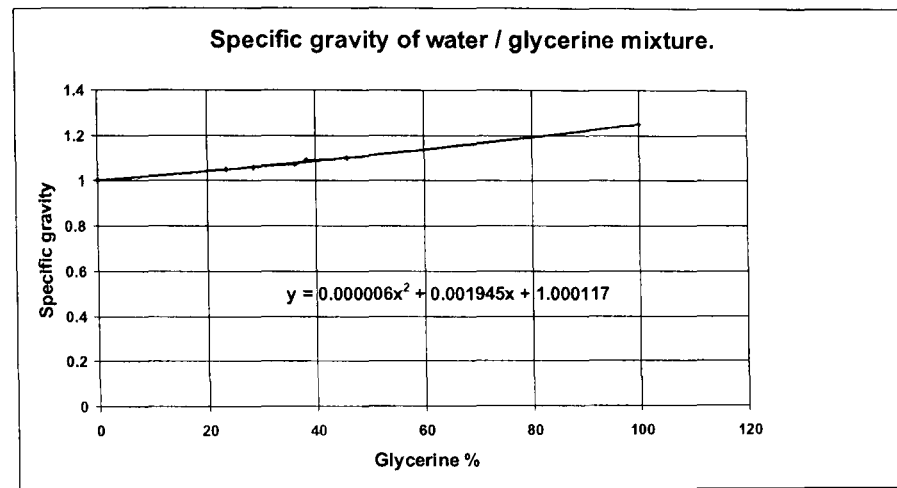
Test sheet 18 of rotor 0

Appendix 12.

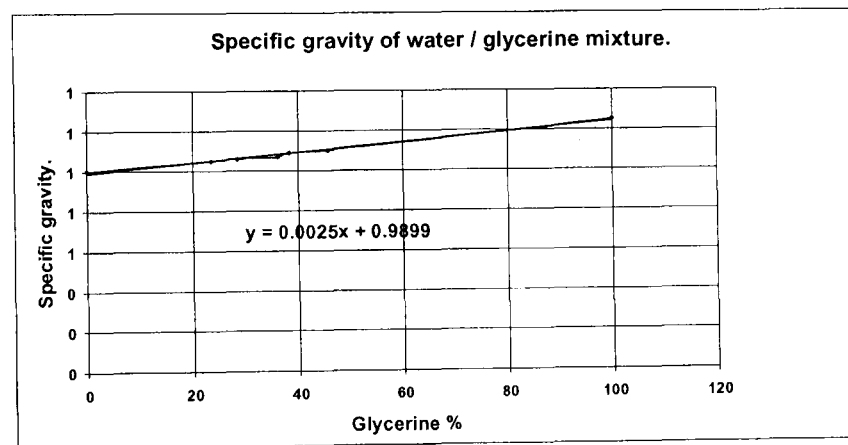
The performance 'set points' for the constant speed tests.

Delivery l, min ⁻¹	Speed rev.min ⁻¹
10	4000
10	3500
10	3000
10	2500
10	2000
10	1500
10	1000
5	4000
5	3500
5	3000
5	2500
5	2000
5	1500
5	1000

Appendix 13.1. For a 45% glycerine in water mixture, the polyline result gives a specific gravity of 1.099792.

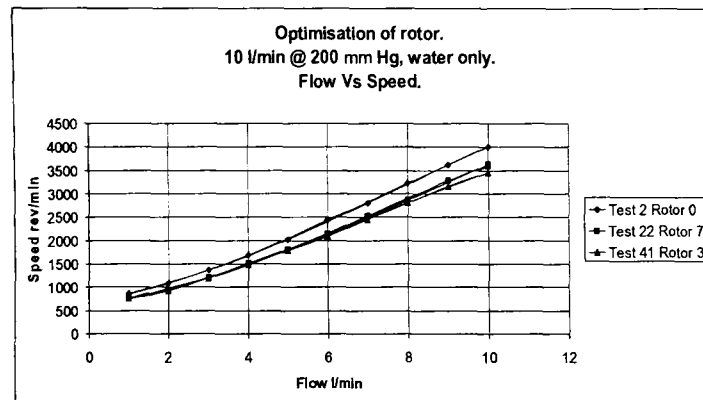


Appendix 13.2. For a 45% glycerine in water mixture the straight line result gives a specific gravity of 1.1024



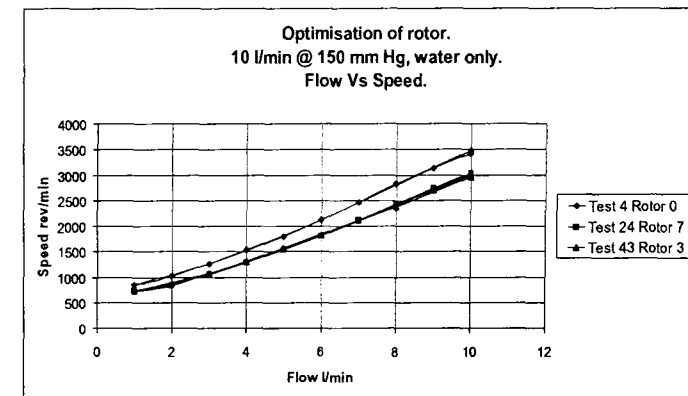
Appendix 14.1. Test fluid, water only.

Optimisation of rotor, 10 l/min @ 200 mm Hg.



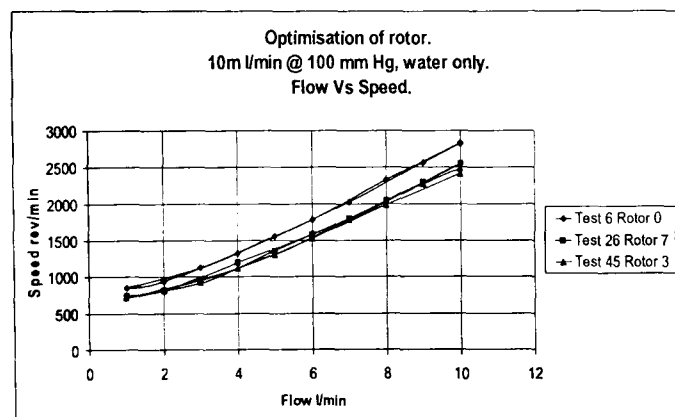
Appendix 14.2. Test fluid, water only

Optimisation of rotor, 10 l/min @ 150 mm Hg



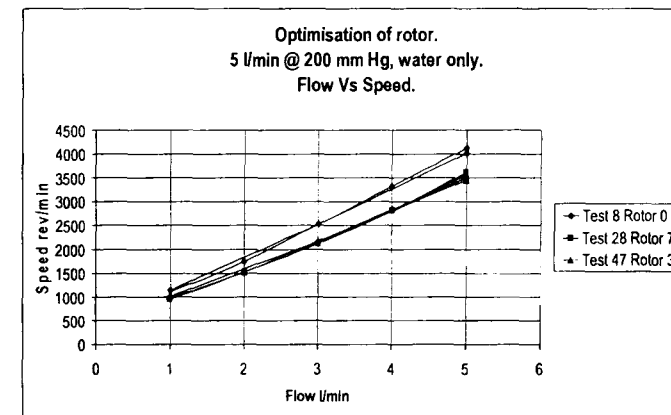
Appendix 14.3. Test fluid, water only

Optimisation of rotor, 10 l/min @ 100 mm Hg.



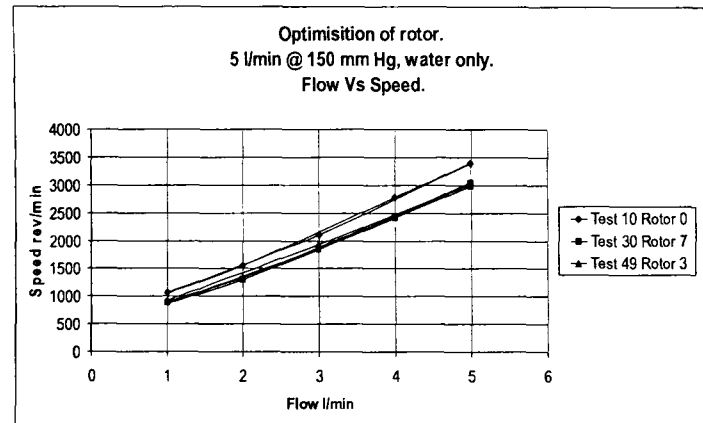
Appendix 14.4. Test fluid, water only

Optimisation of rotor, 5 l/min @ 200 mm Hg.



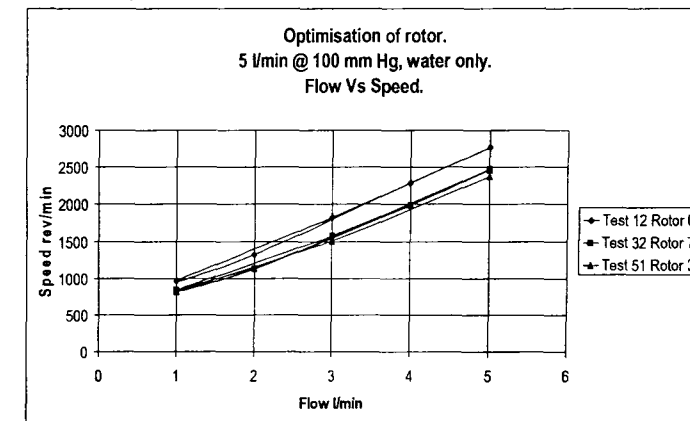
Appendix 14.5. Test fluid, water only

Optimisation of rotor, 5 l/min @ 150 mm Hg.



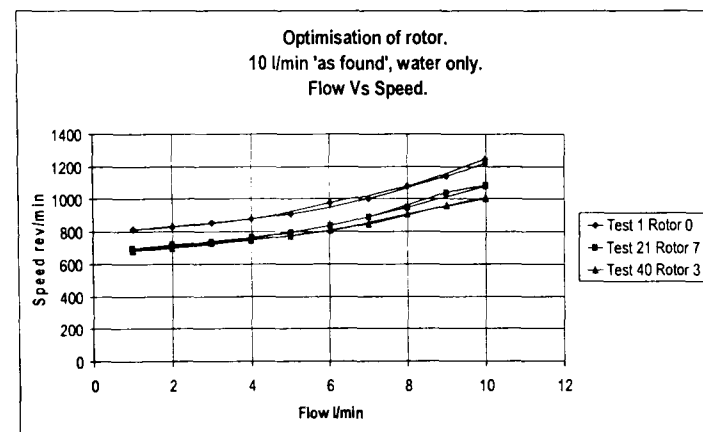
Appendix 14.6. Test fluid, water only.

Optimisation of rotor, 5 l/min @ 100 mm Hg.



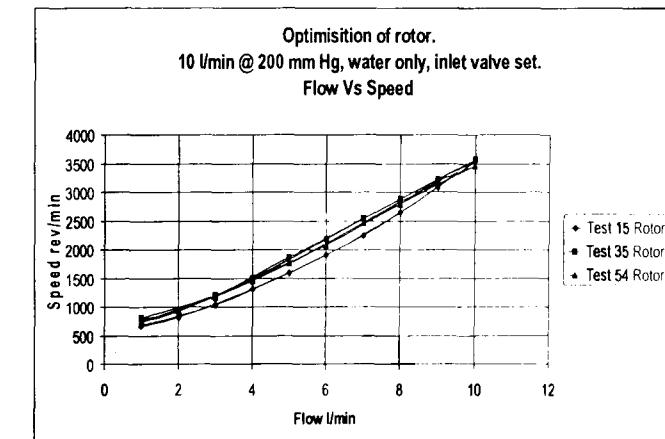
Appendix 14.7. Test fluid, water only.

Optimisation of rotor, 10 l/min, 'as found'.



Appendix 14.8 .Test fluids, water only.

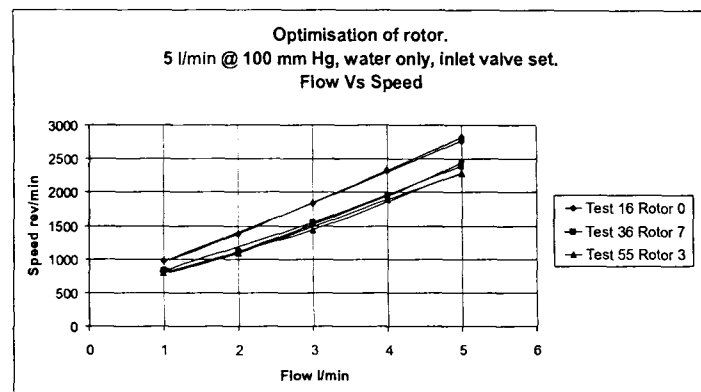
Optimisation of rotor, 10 l/min @ 200 mm Hg. Pump inlet restricted



Appendix 14.9 .Test fluids, water only.

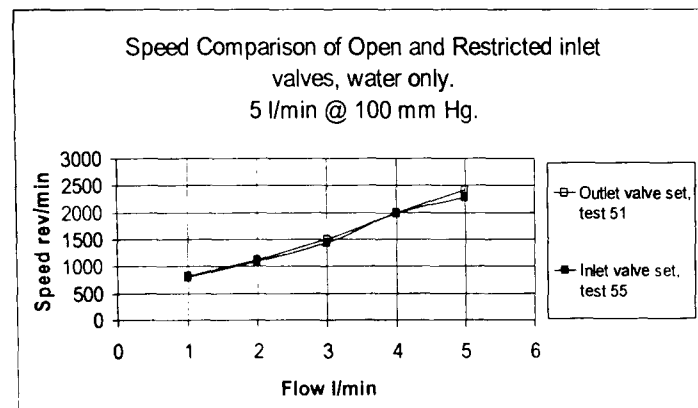
Optimisation of rotor, 5 l/min @ 100 mm Hg.

Pump inlet restricted



Appendix 14.11. Test fluid water only. Comparison of speed Vs flow with inlet valve open and restricted.

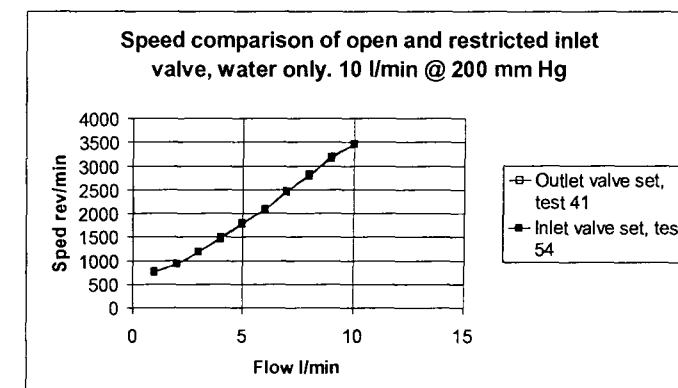
5 l/min @ 100 mm Hg



Appendix 14.10. Test fluid, water only. Comparison

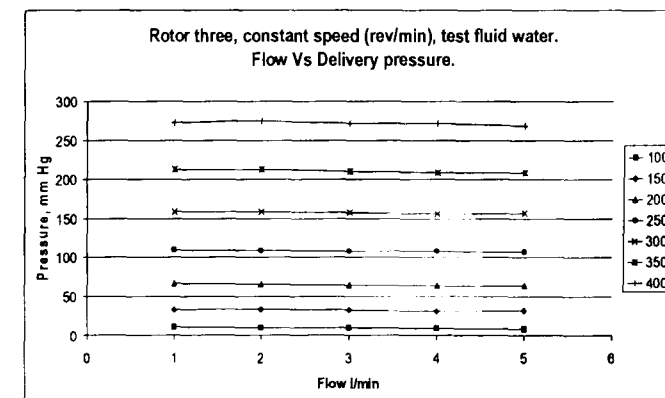
of speed Vs flow with inlet valve open and restricted.

10 l/min @ 200 mm Hg.

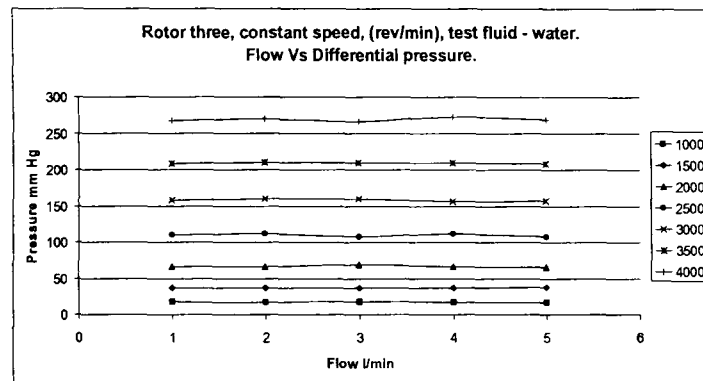


Appendix 14.12. Test fluid – water only. Constant speed tests, 1000 – 4000 rev/min. Flow Vs delivery pressure

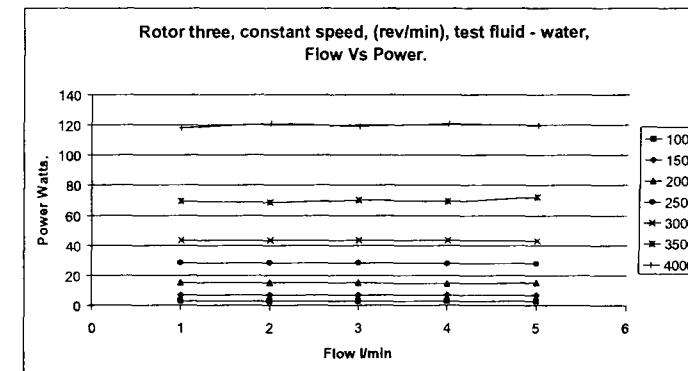
with optimum rotor @ 5 l/min



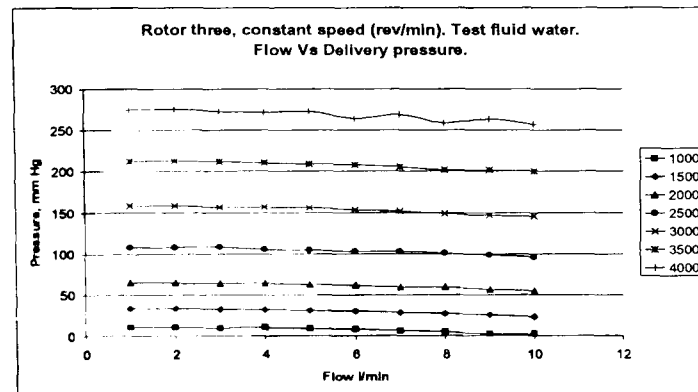
Appendix 14.13. Test fluid – water. Constant speed.
tests, 1000 – 4000 rev/min. Flow Vs differential pressure
with optimum rotor @ 5 l/min.



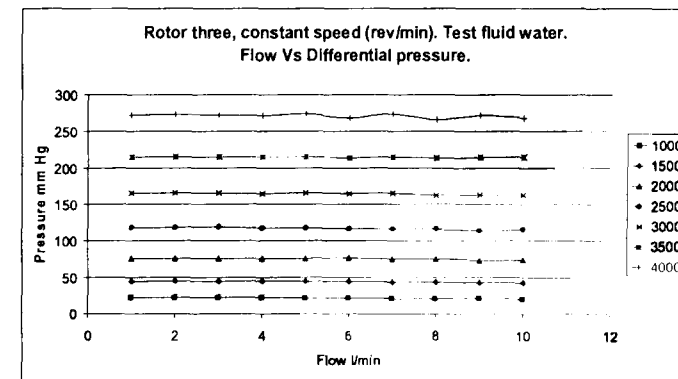
Appendix 14.14. Test fluid – water. Constant speed,
tests 1000 – 4000 rev/min. Flow Vs power
with optimum rotor @ 5 l/min.



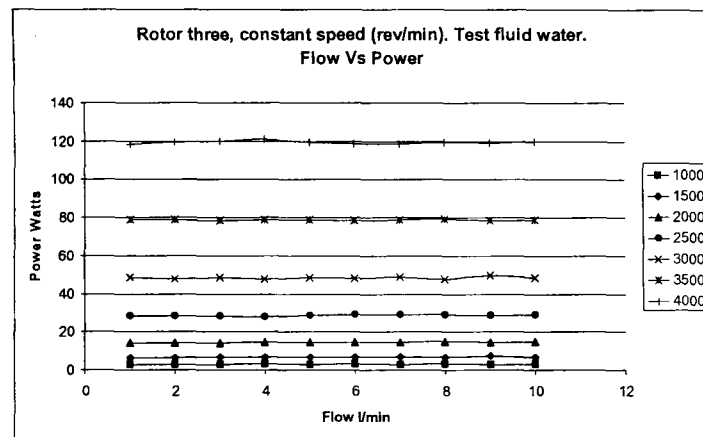
Appendix 14.15. Test fluid – water. Constant speed
tests, 1000 – 4000 rev/min., Flow Vs delivery pressure
with optimum rotor @ 10 l/min



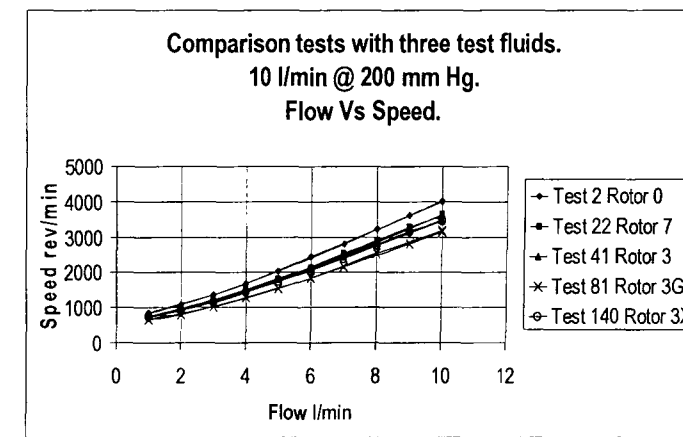
Appendix 14.16. Test fluid – water. Constant speed
tests, 1000 – 4000 rev/min Flow Vs differential pressure
with optimum rotor @ 10 l/min



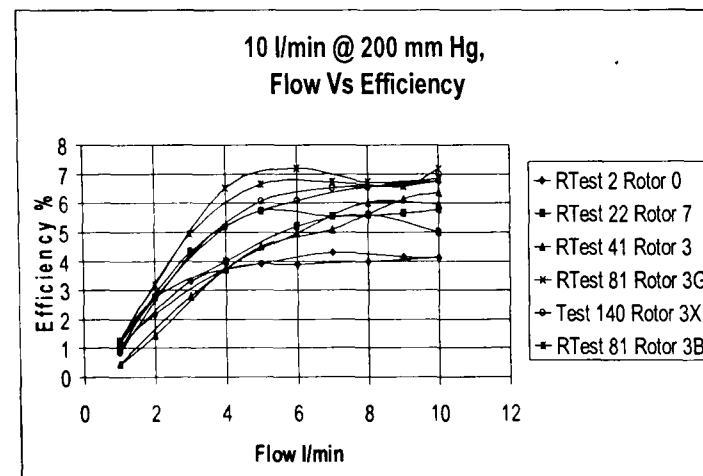
Appendix 14.17. Test fluid – water. Constant speed tests, 1000 – 4000 rev/min. Flow Vs power with optimum rotor @ 10 l/min.



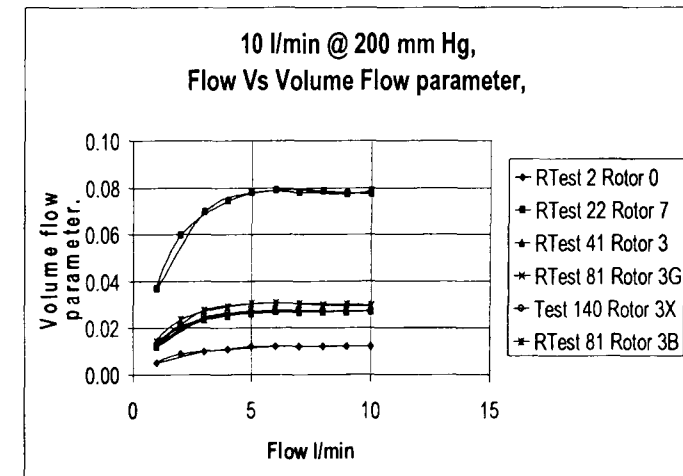
Appendix 14.18. Comparison of flow Vs speed with all test fluids.



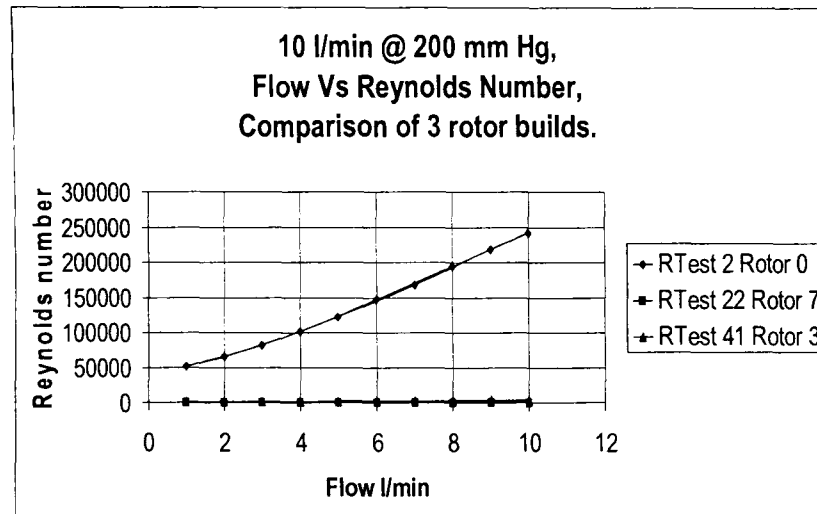
Appendix 14.19 – Flow Vs Efficiency



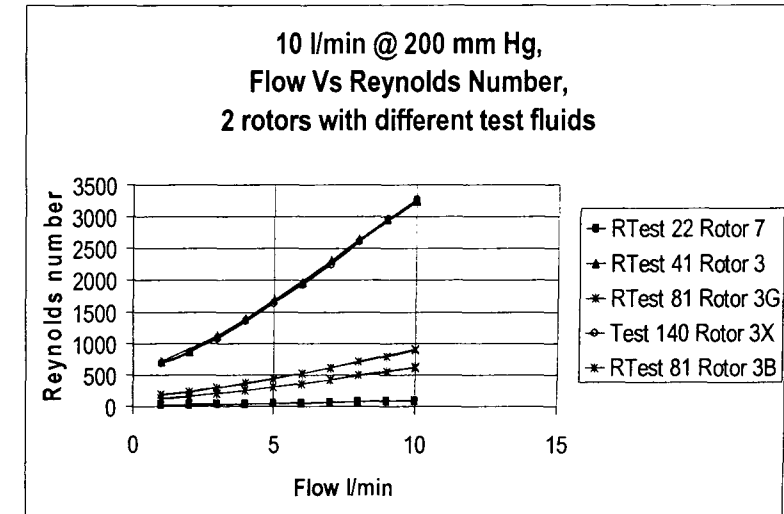
Appendix 14.20 – Flow Vs Flow Parameter.



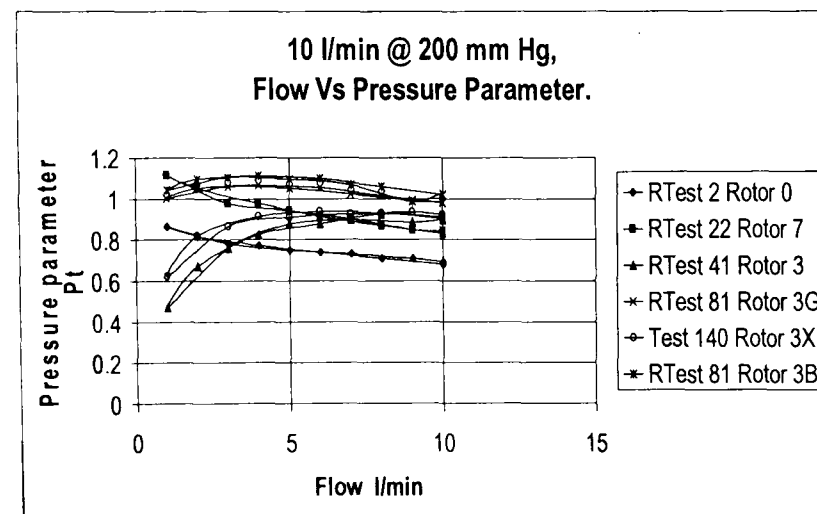
Appendix 14.21 – Flow Vs Reynolds Number



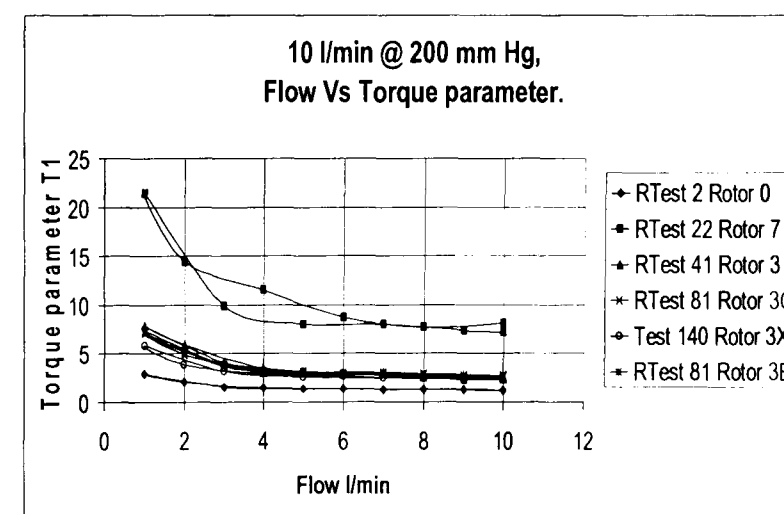
Appendix 14.22 – Flow Vs Reynolds Number



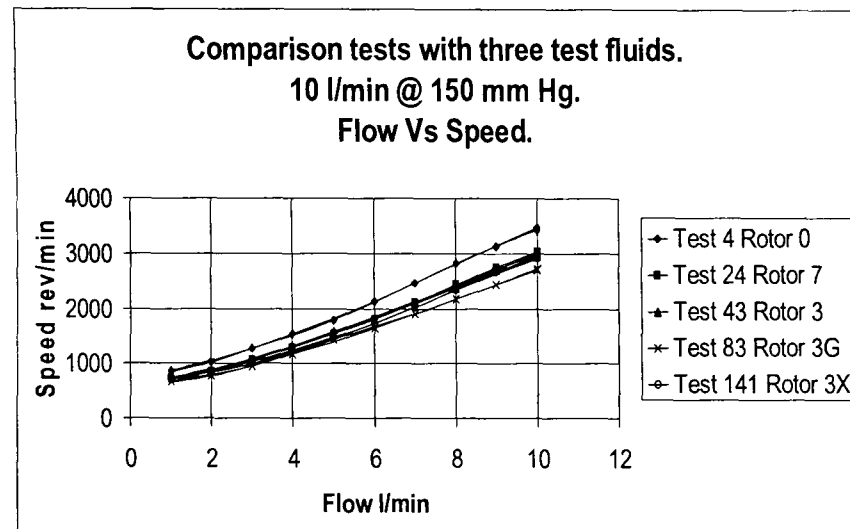
Appendix 14.23 – Flow Vs Pressure Parameter.



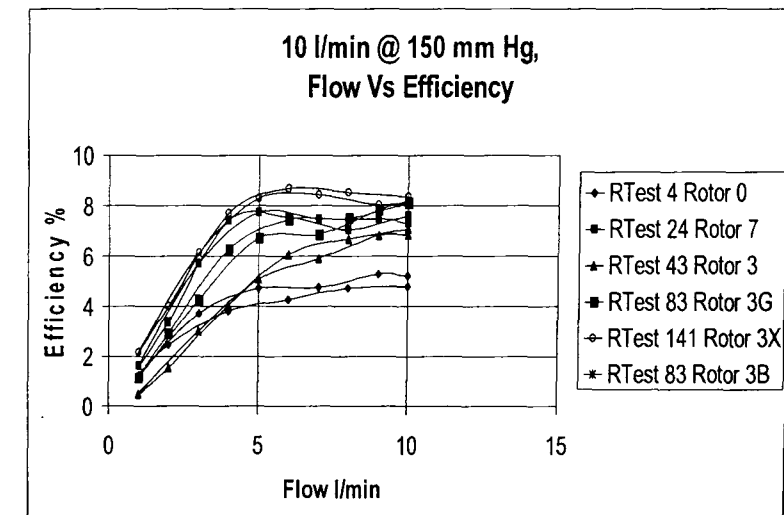
Appendix 14.24 – Flow Vs Torque Parameter.



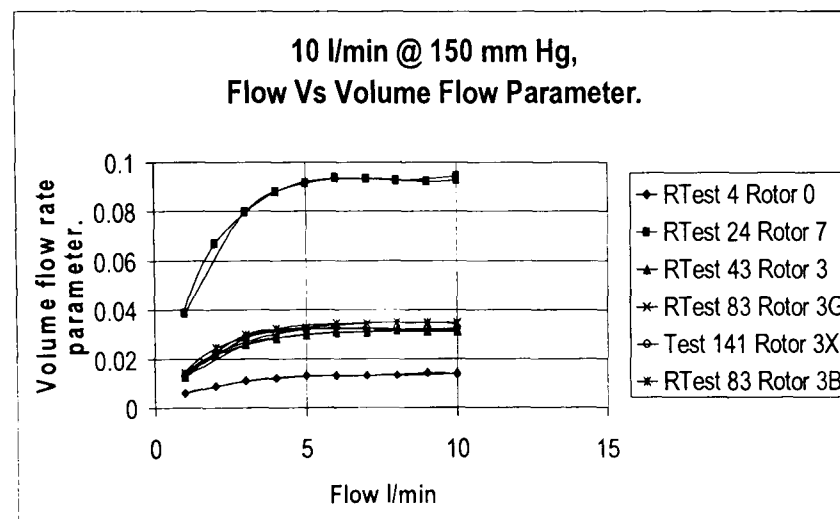
Appendix 14.25 – Flow Vs Speed



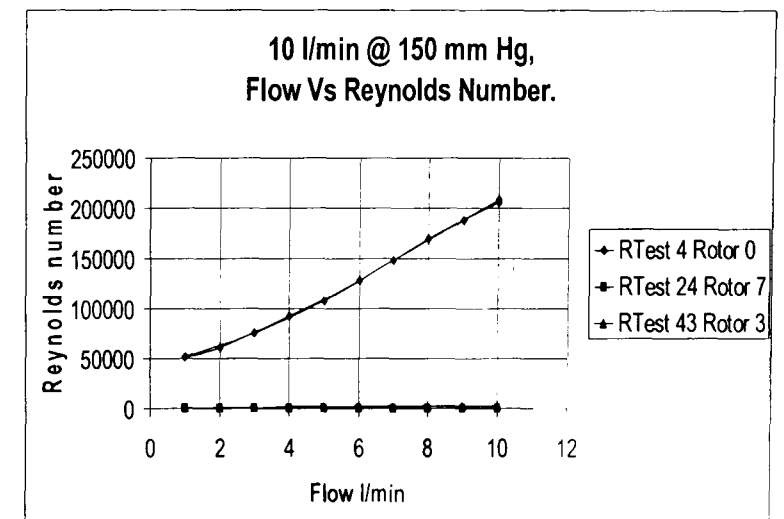
Appendix 14.26 – Flow Vs Efficiency



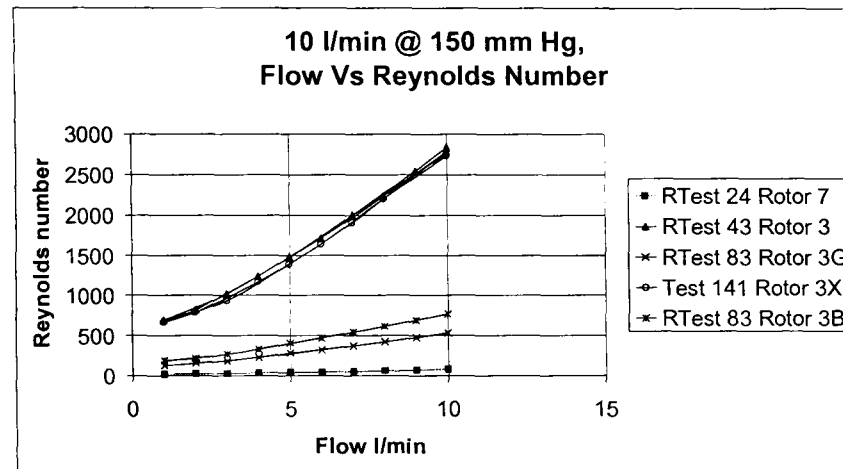
Appendix 14.27 – Flow Vs Flow parameter.



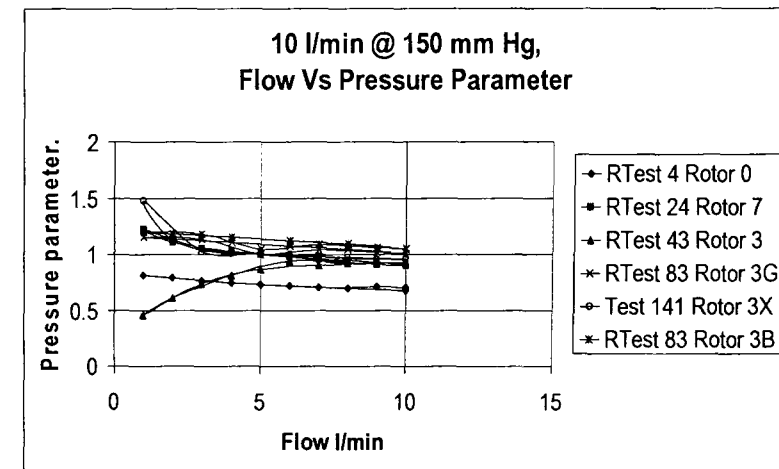
Appendix 14.28 – Flow Vs Reynolds number



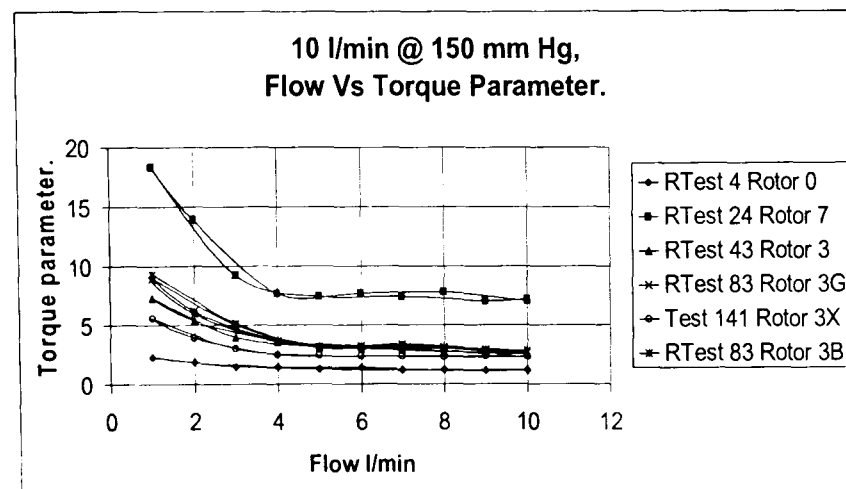
Appendix 14.29 – Flow Vs Reynolds number.



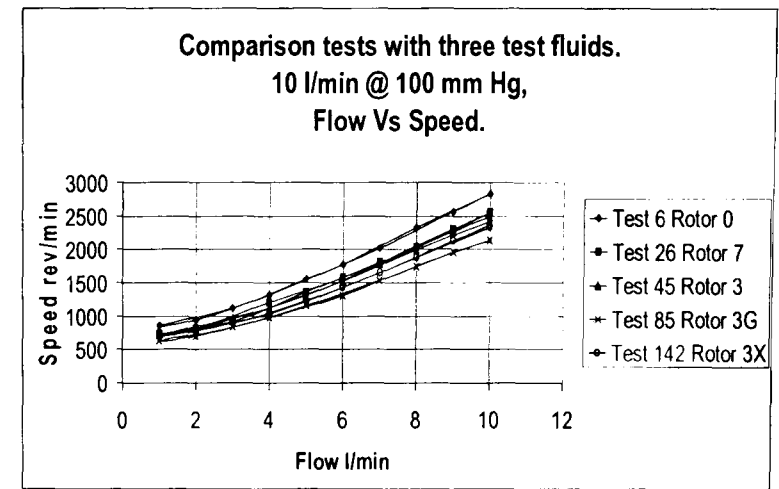
Appendix 14.30 – Flow Vs Pressure parameter



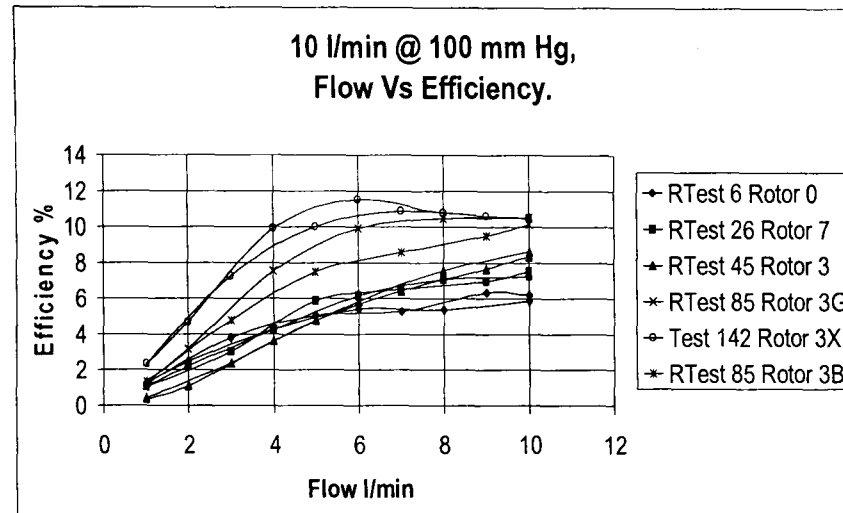
Appendix 14.31 – Flow Vs Torque parameter.



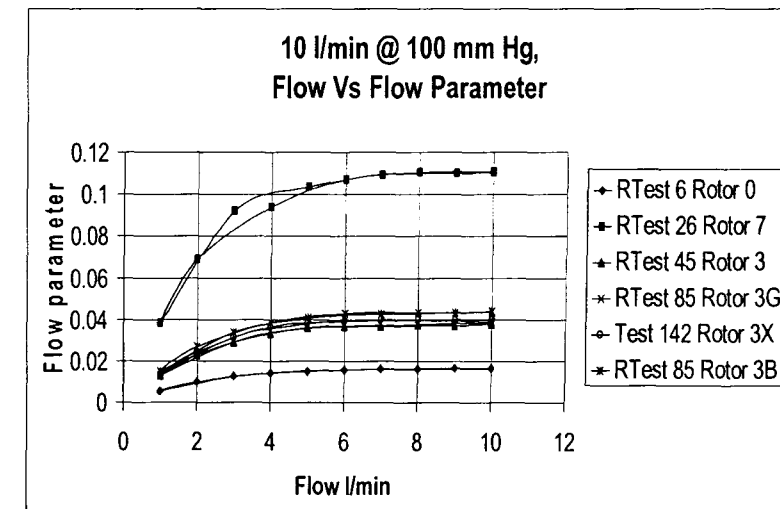
Appendix 14.32 – Flow Vs Speed.



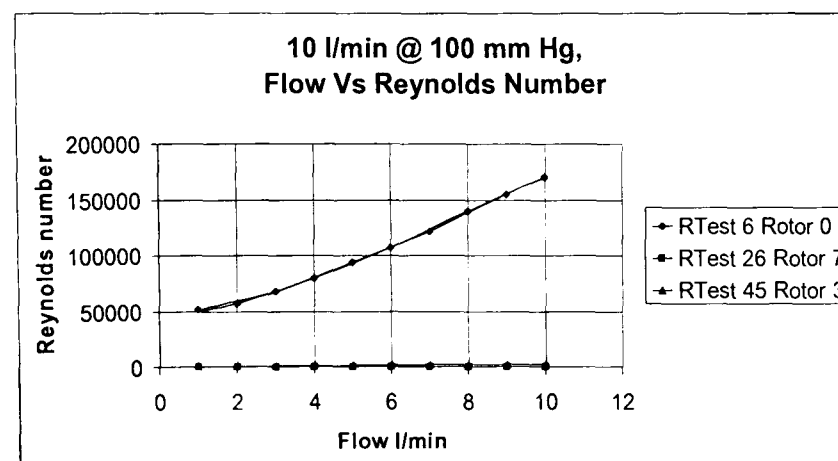
Appendix 14.33 – Flow Vs Efficiency.



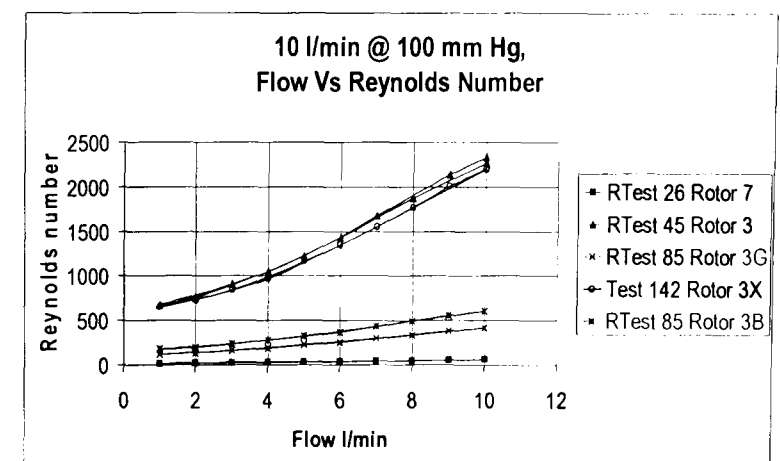
Appendix 14.34 – Flow Vs Flow parameter.



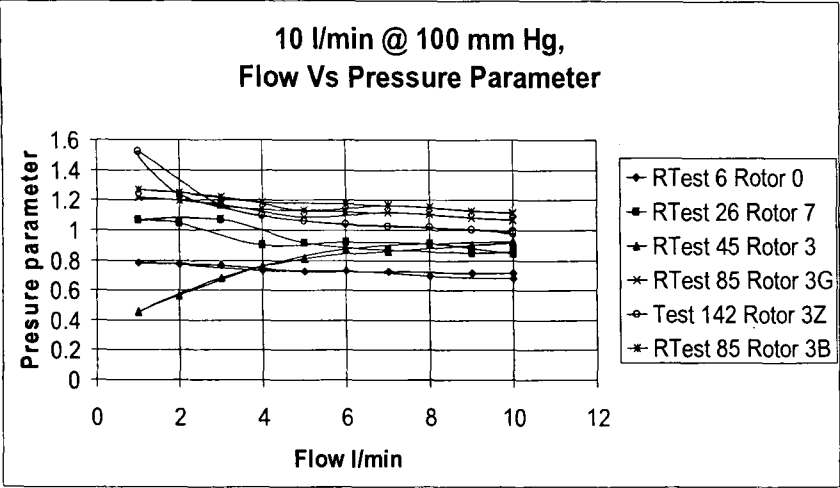
Appendix 14.35 – Flow Vs Reynolds number



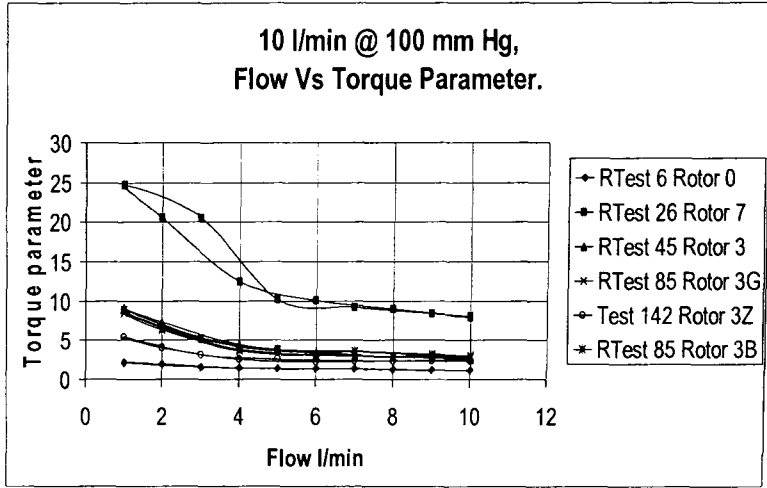
Appendix 14.36 – Flow Vs Reynolds number



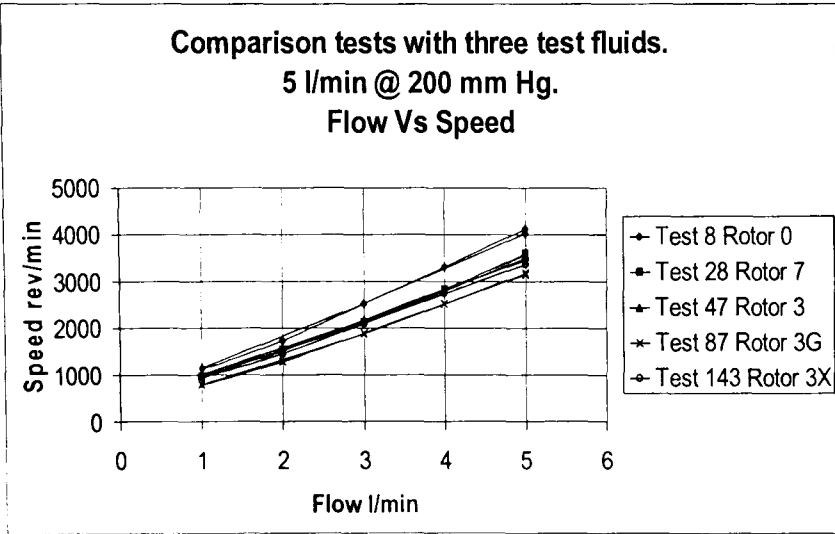
Appendix 14.37 – Flow Vs Pressure parameter



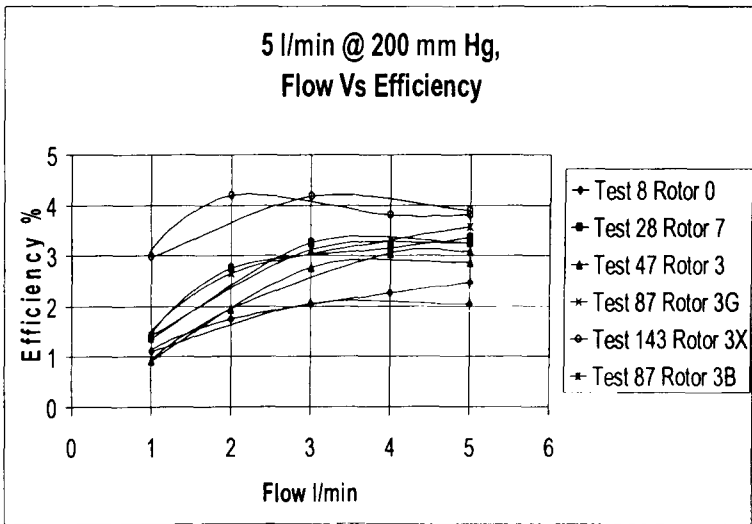
Appendix 14.38– Flow Vs Torque parameter.



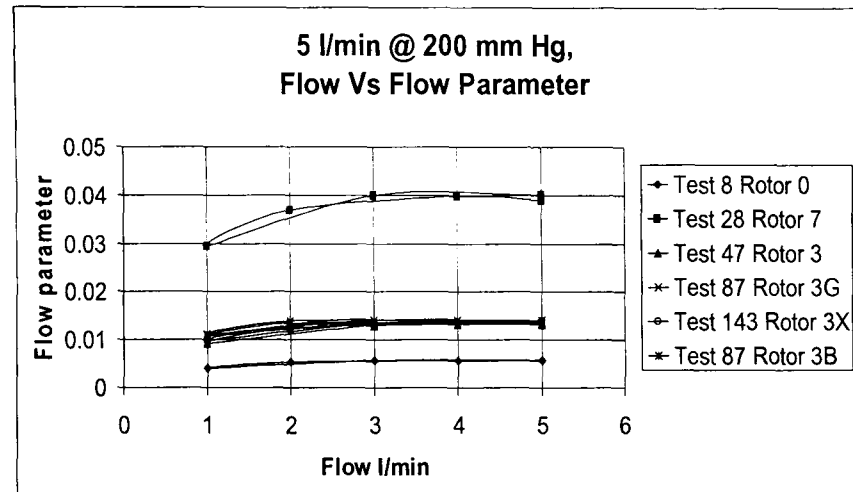
Appendix 14.39 – Flow Vs Speed.



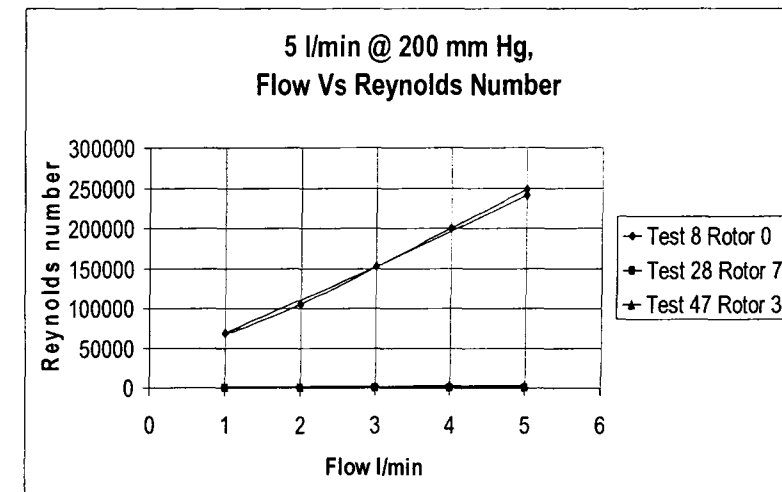
Appendix 14.40 – Flow Vs Efficiency.



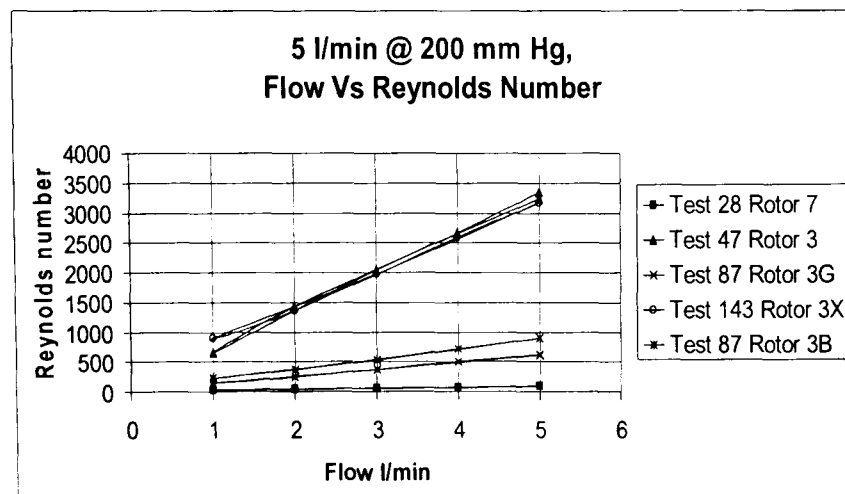
Appendix 14.41 – Flow Vs Flow parameter.



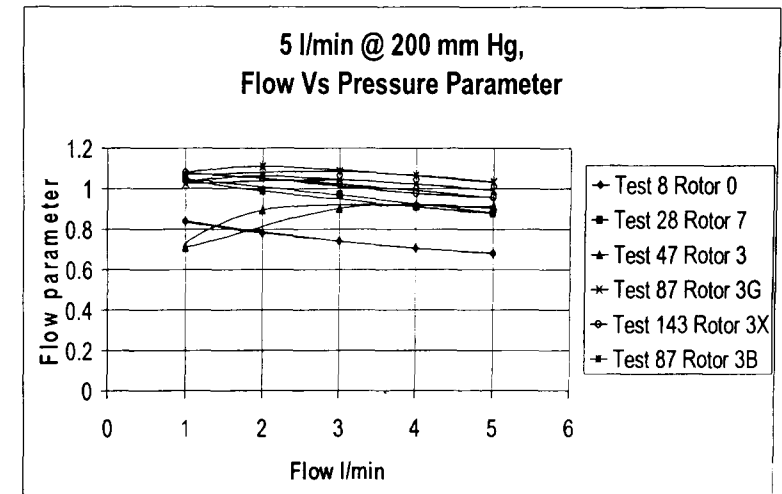
Appendix 14.42 – Flow Vs Reynolds number



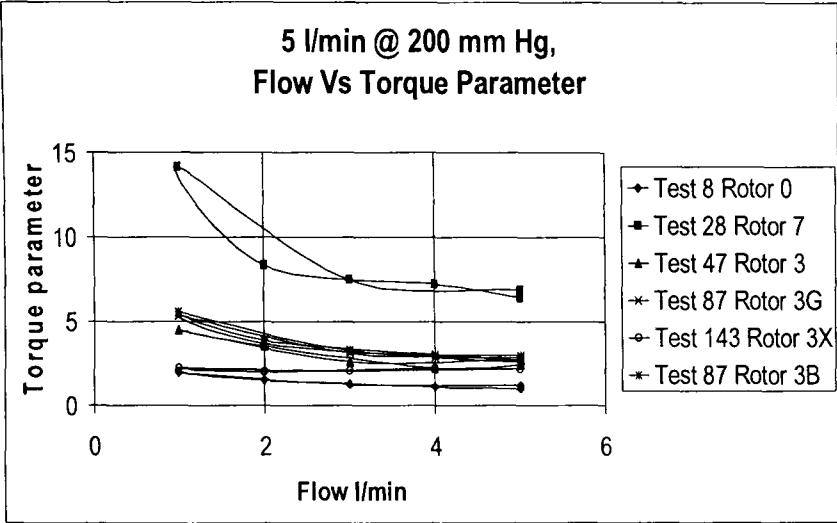
Appendix 14.43– Flow Vs Reynolds number



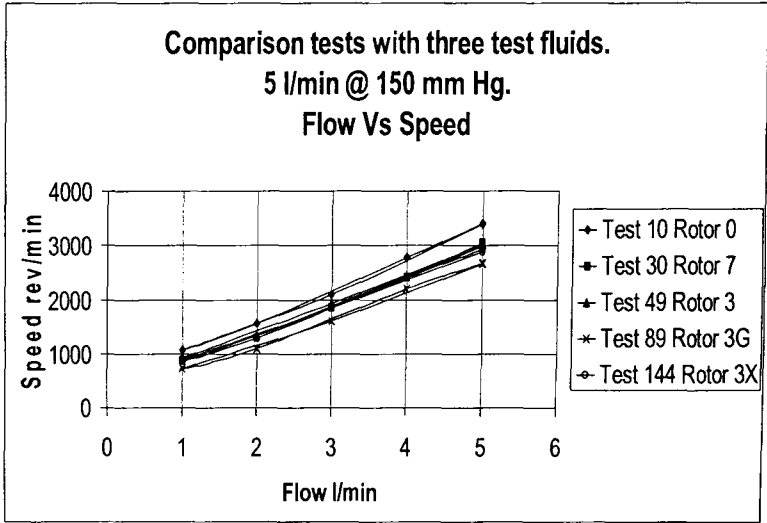
Appendix 14.44 – Flow Vs Pressure parameter



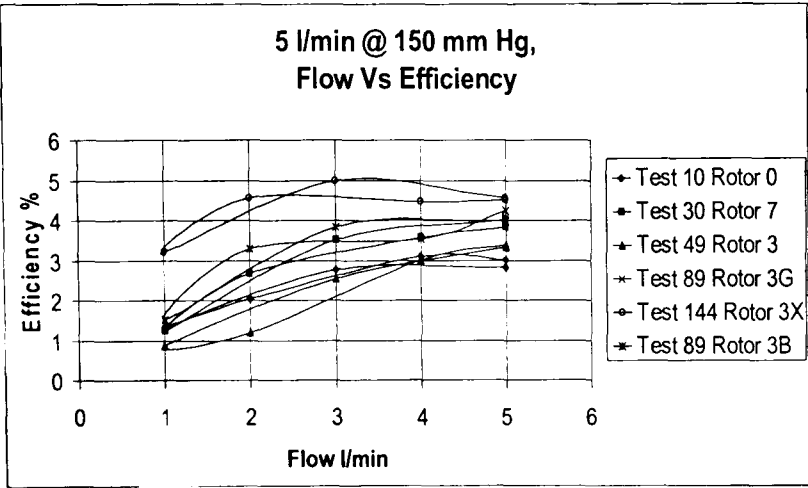
Appendix 14.45 – Flow Vs Torque parameter.



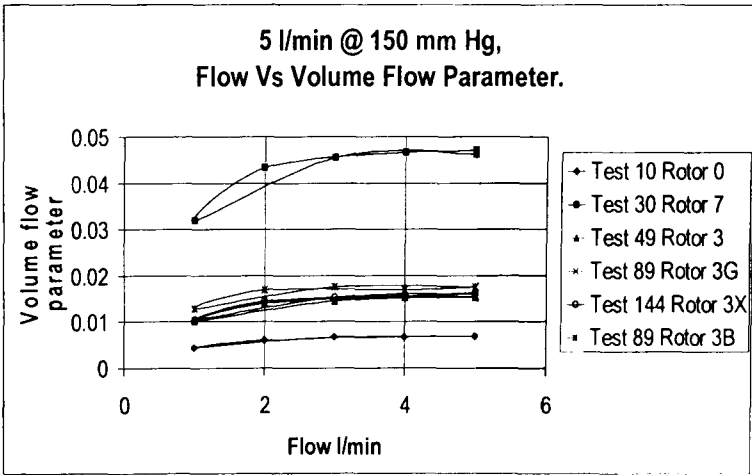
Appendix 14.46 – Flow Vs Speed.



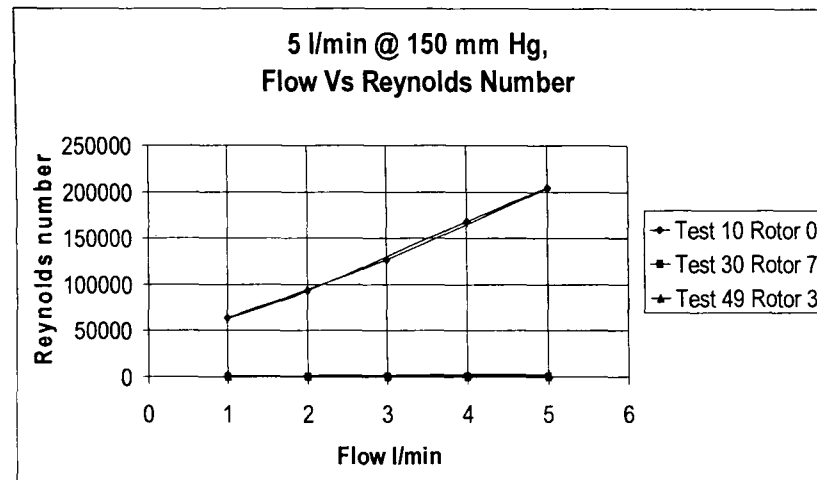
Appendix 14.47 – Flow Vs Efficiency.



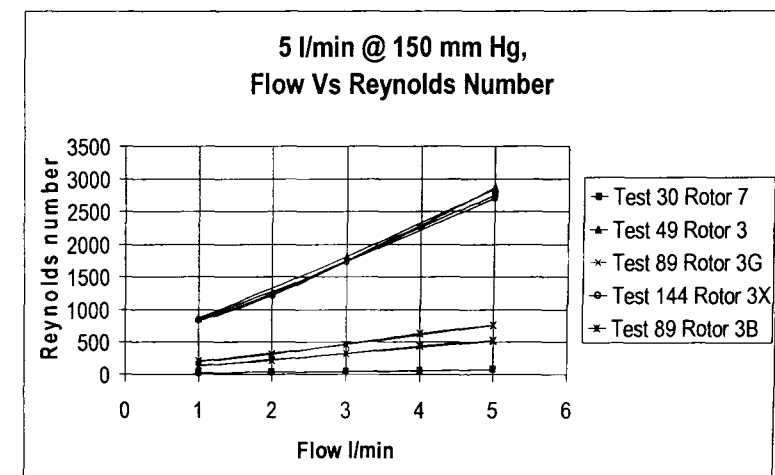
Appendix 14.48 – Flow Vs Flow parameter



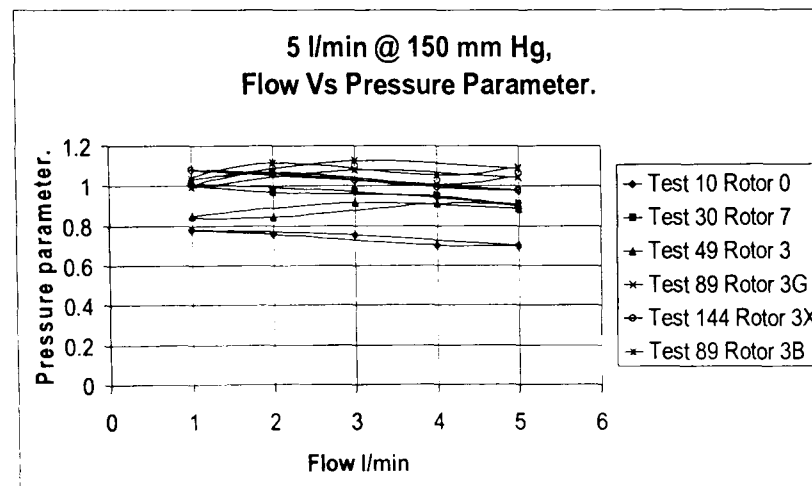
Appendix 14.49 – Flow Vs Reynolds number



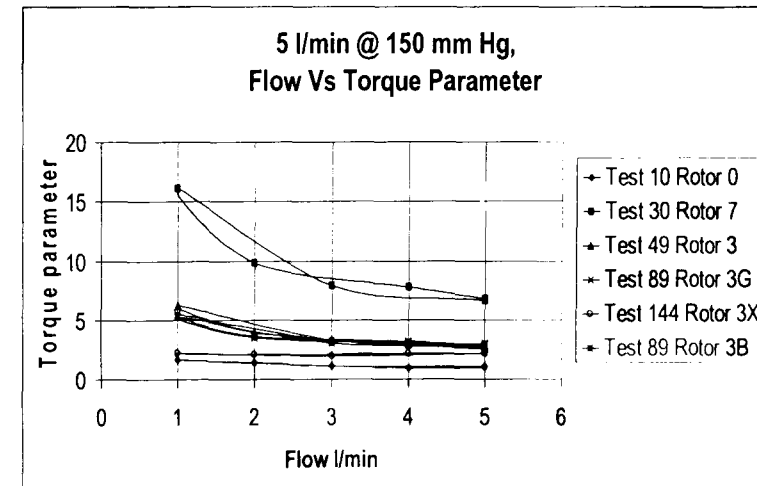
Appendix 14.50 – Flow Vs Reynolds number



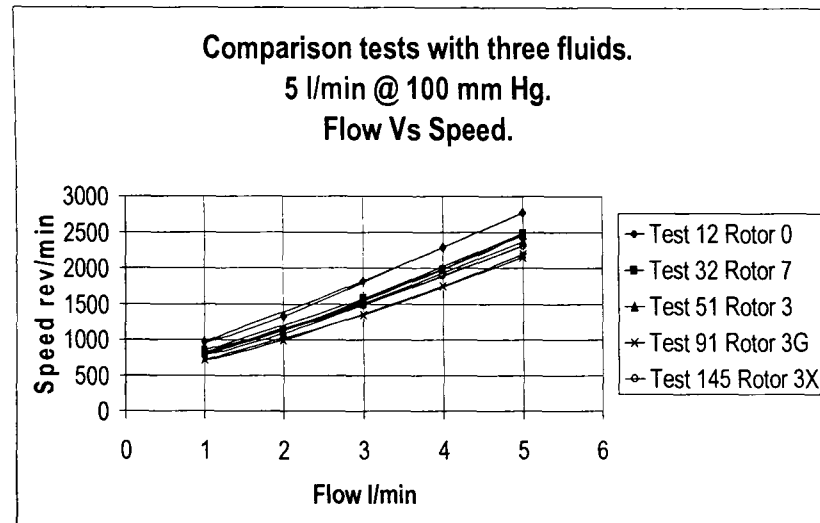
Appendix 14.51 – Flow Vs Pressure parameter



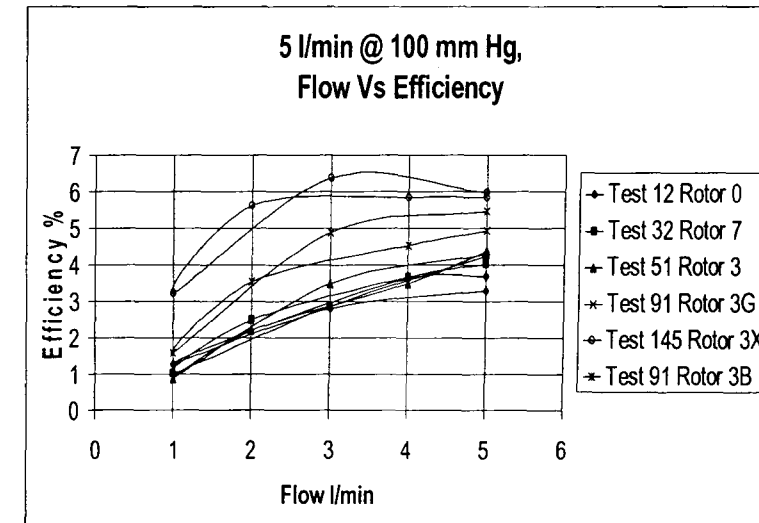
Appendix 14.52 – Flow Vs Torque parameter.



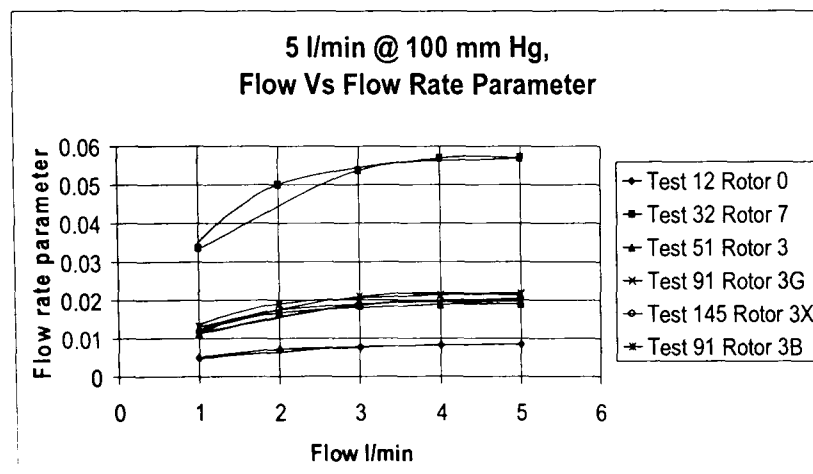
Appendix 14.53 – Flow Vs Speed.



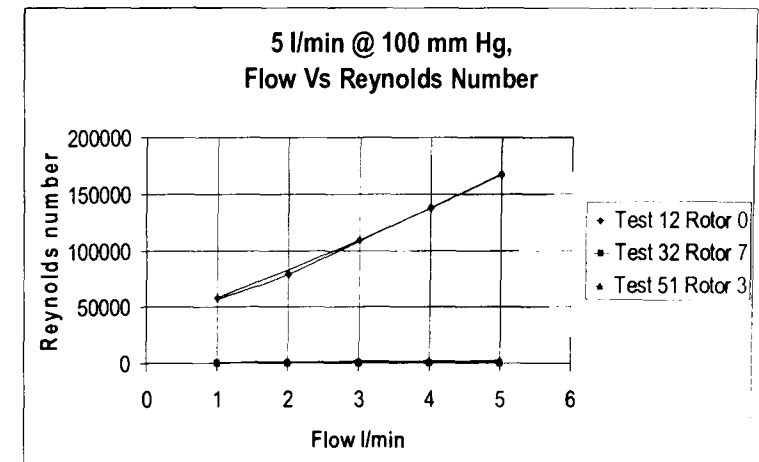
Appendix 14.54 – Flow Vs Efficiency



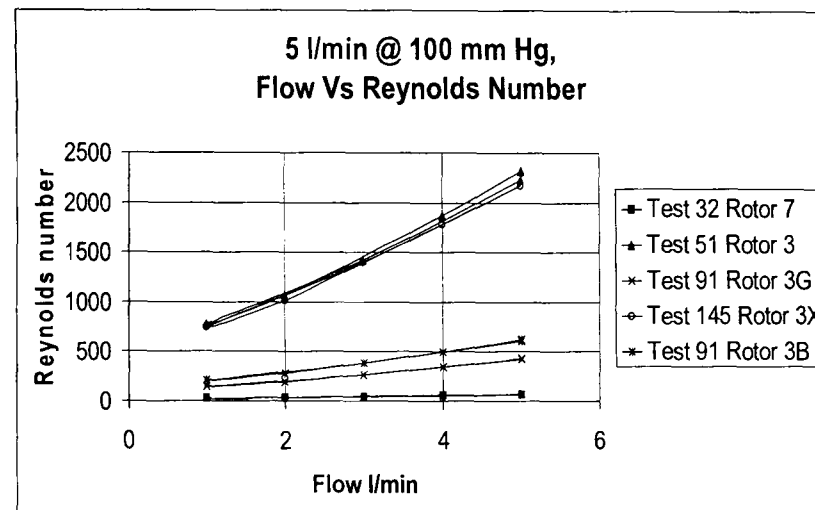
Appendix 14.55 – Flow Vs Flow parameter



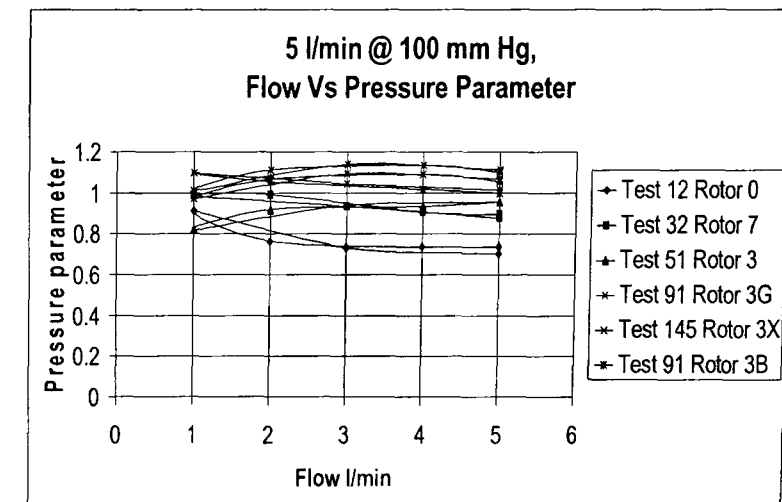
Appendix 14.56 – Flow Vs Reynolds number



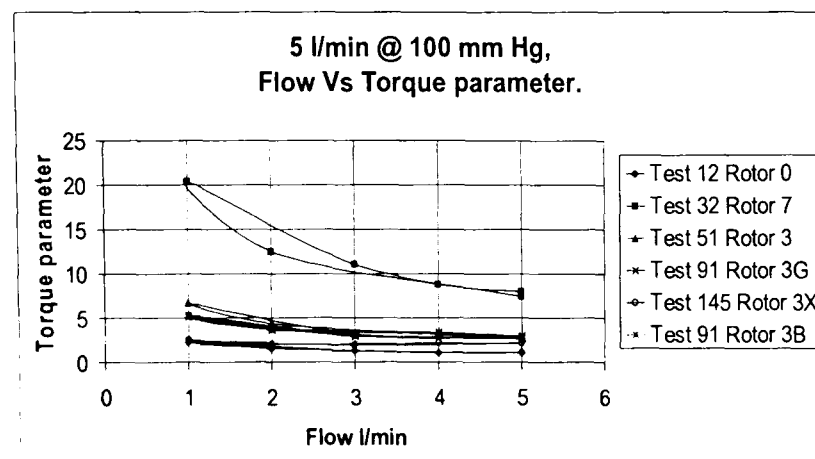
Appendix 14.57 – Flow Vs Reynolds number



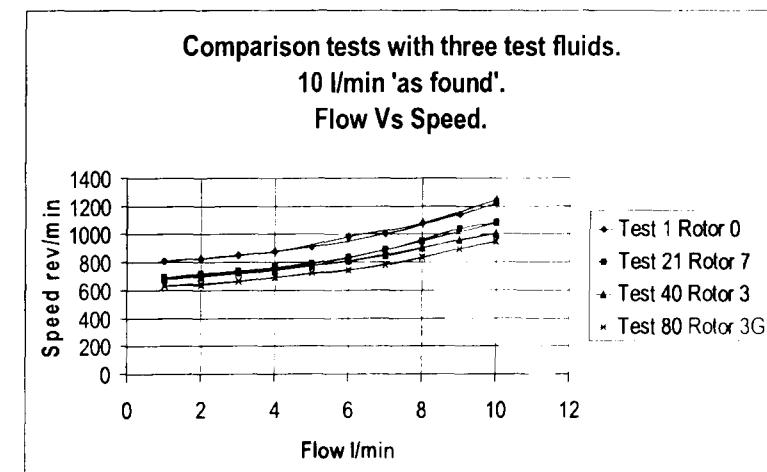
Appendix 14.58 – Flow Vs Pressure parameter



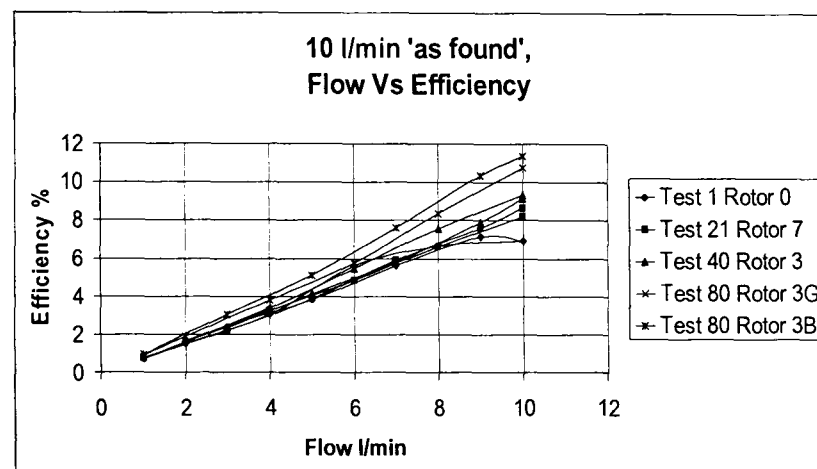
Appendix 14.59 – Flow Vs Torque parameter.



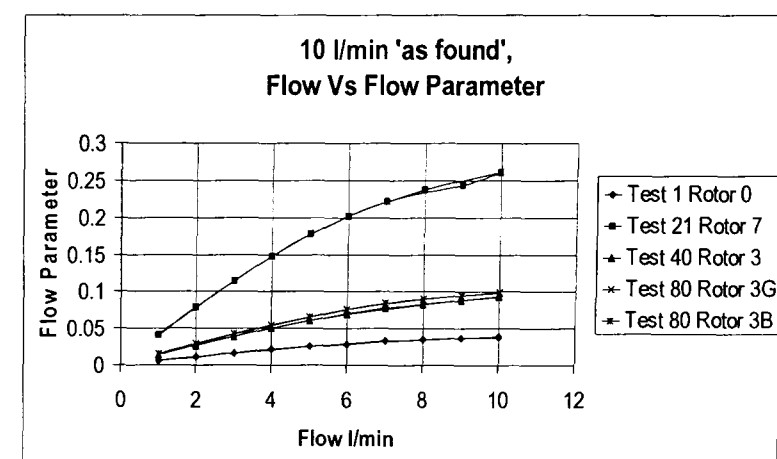
Appendix 14.60 – Flow Vs Speed



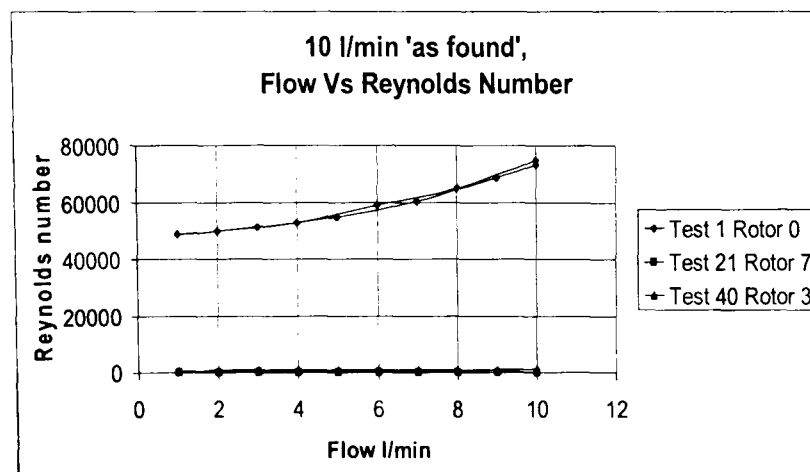
Appendix 14.61 – Flow Vs Efficiency.



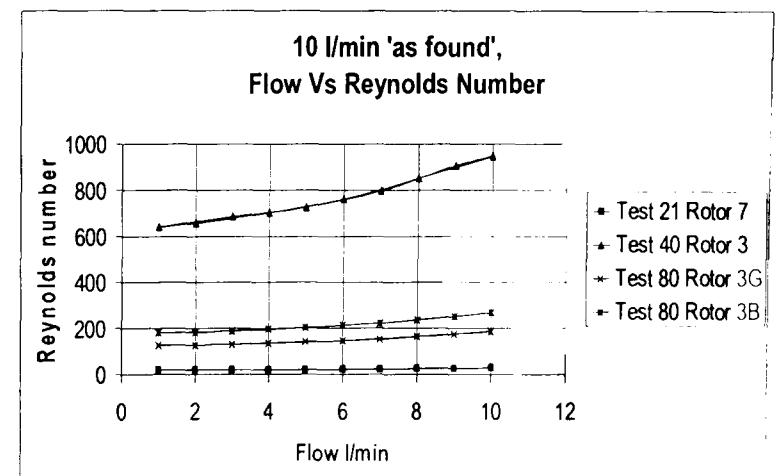
Appendix 14.62 – Flow Vs Flow parameter



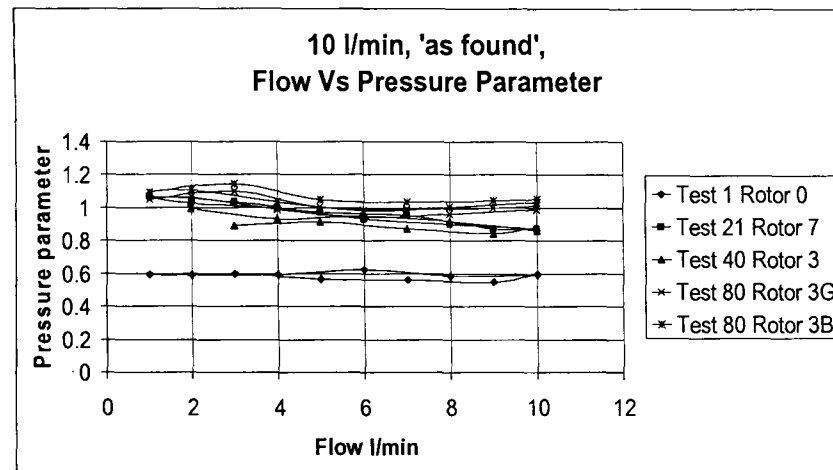
Appendix 14.63 – Flow Vs Reynolds number



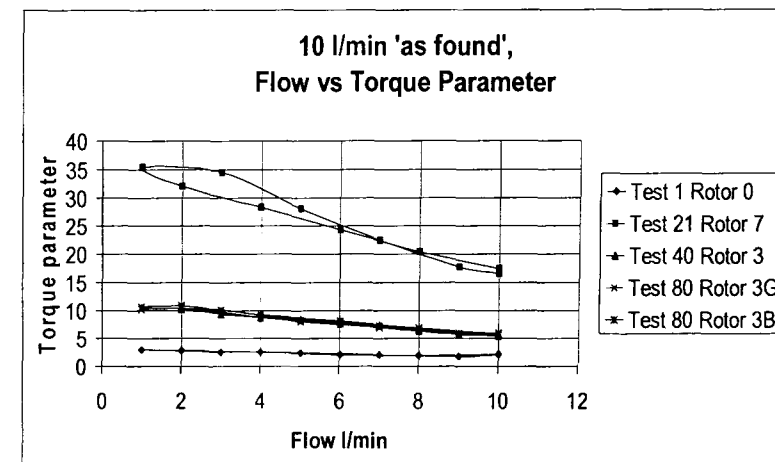
Appendix 14.64 – Flow Vs Reynolds number



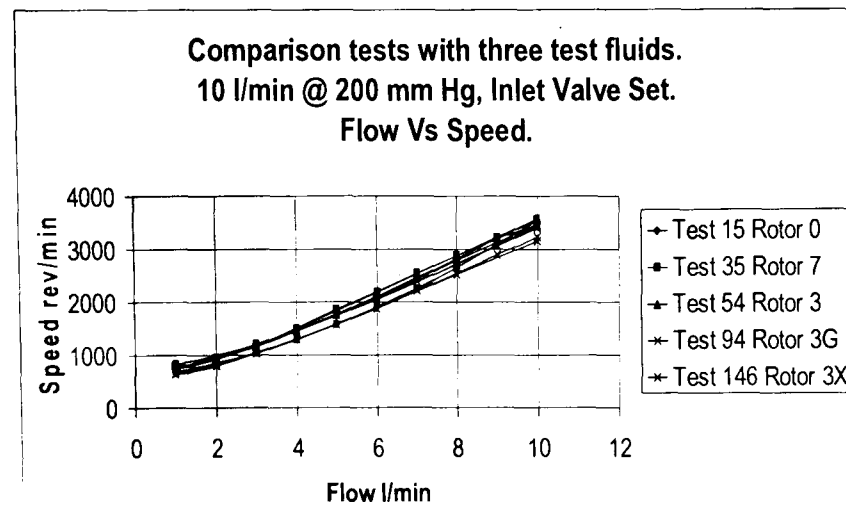
Appendix 14.65 – Flow Vs Pressure parameter



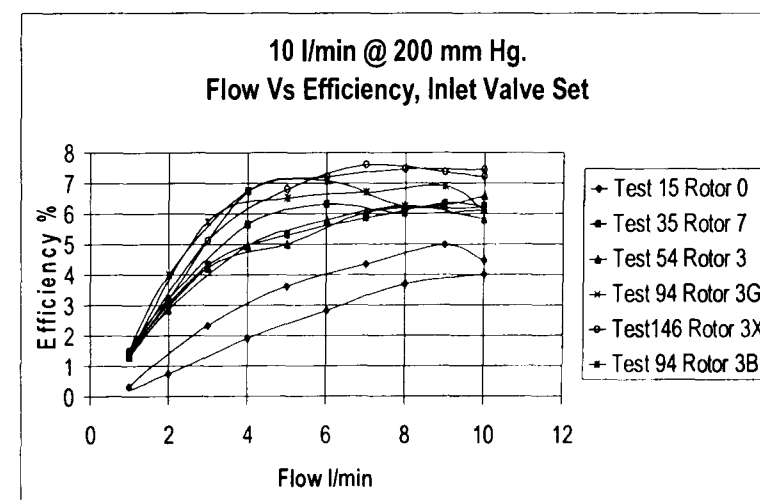
Appendix 14.66 – Flow Vs Torque parameter



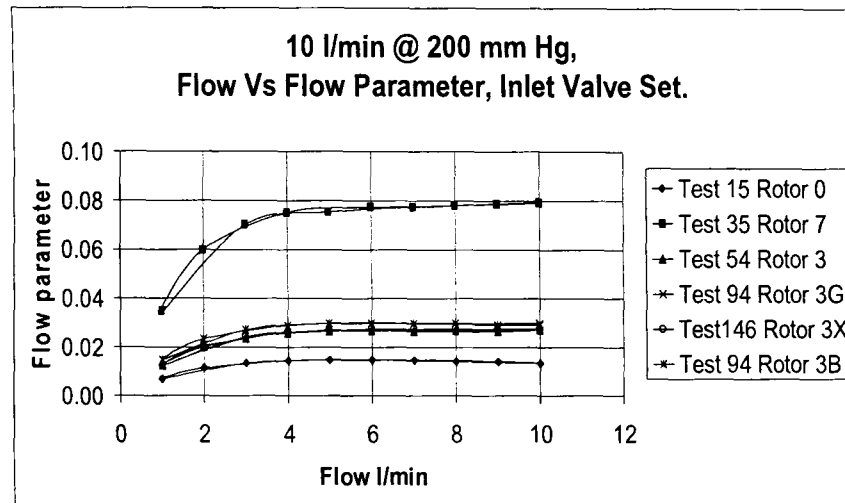
Appendix 14.67– Flow Vs Speed.



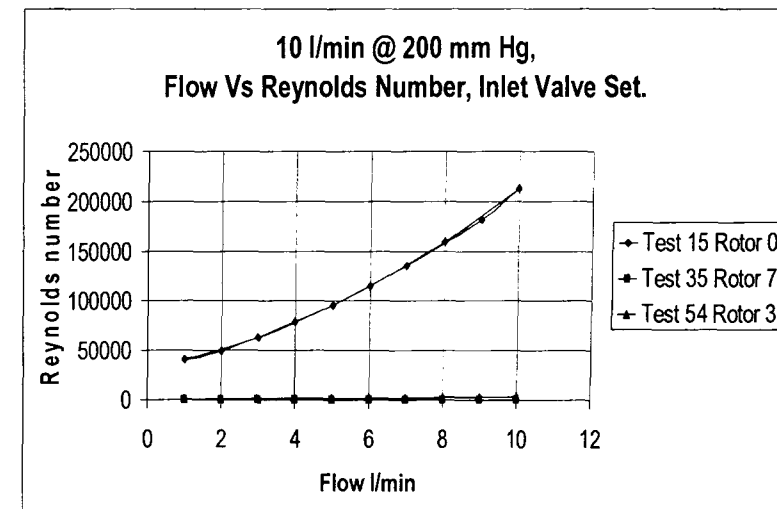
Appendix 14.68 – Flow Vs Efficiency.



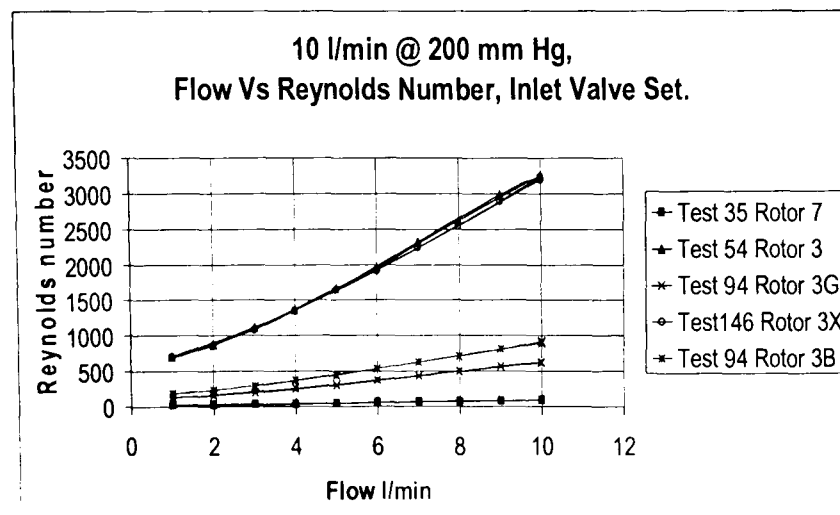
Appendix 14.69 – Flow Vs Flow parameter



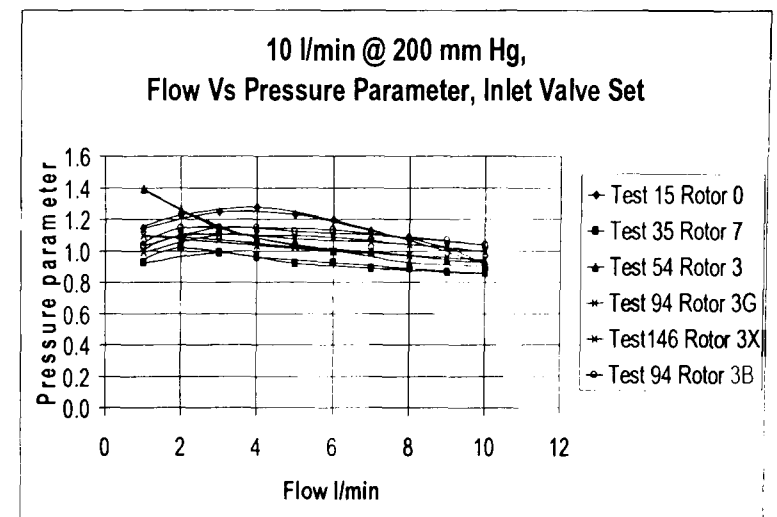
Appendix 14.70 – Flow Vs Reynolds number



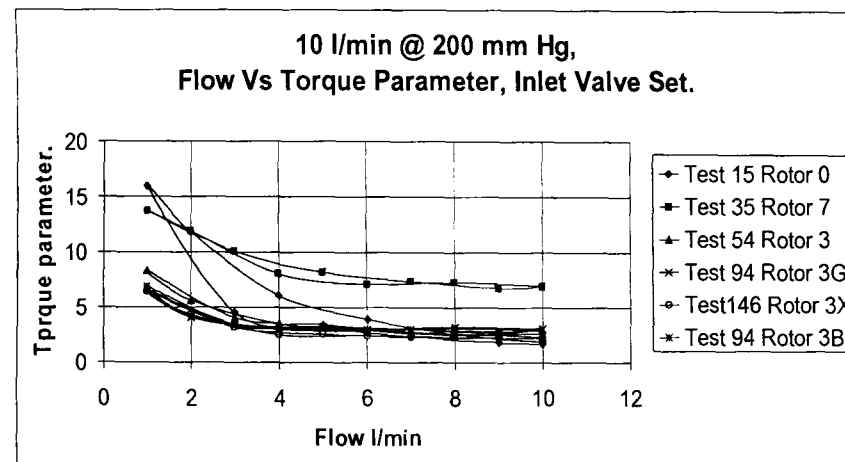
Appendix 14.71 – Flow Vs Reynolds number



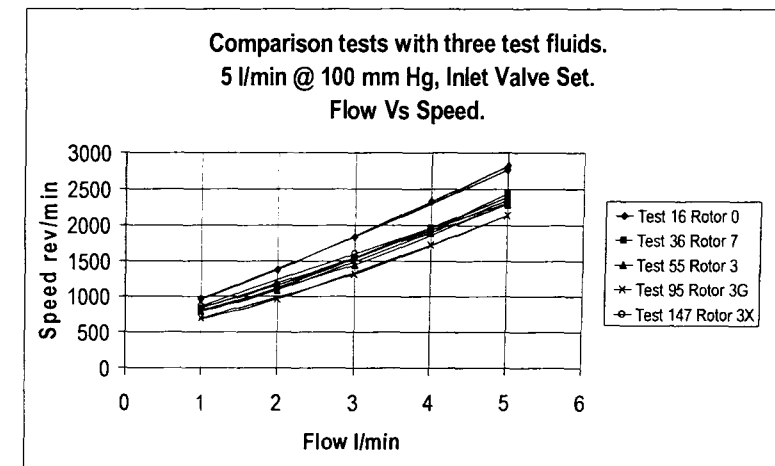
Appendix 14.72 – Flow Vs Pressure parameter



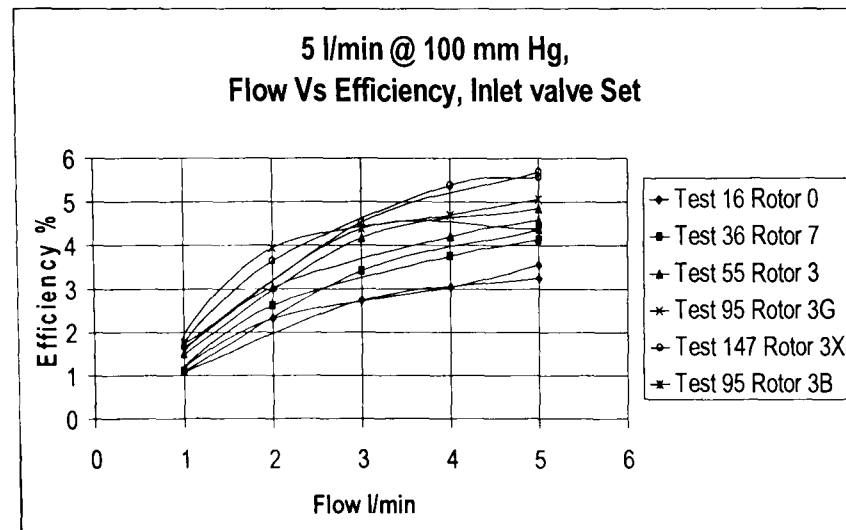
Appendix 14.73 – Flow Vs Torque parameter.



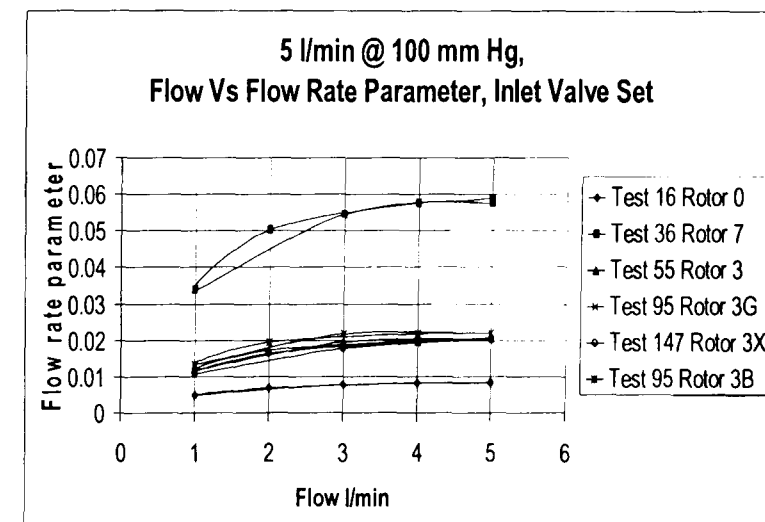
Appendix 14.74 – Flow Vs Speed.



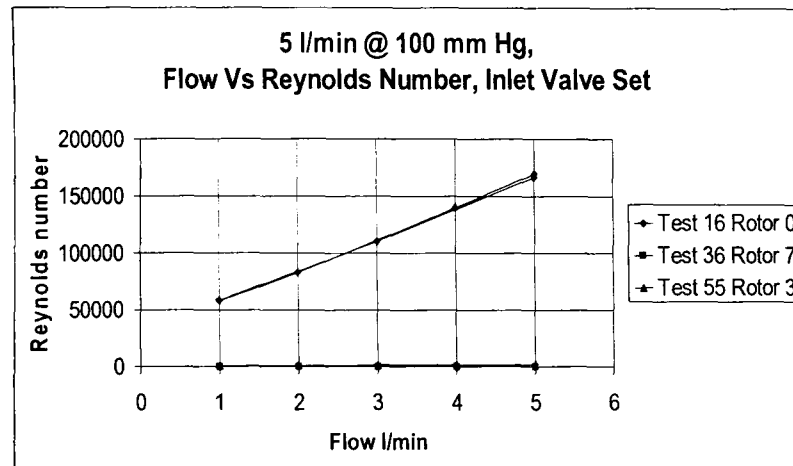
Appendix 14.75 – Flow Vs Efficiency.



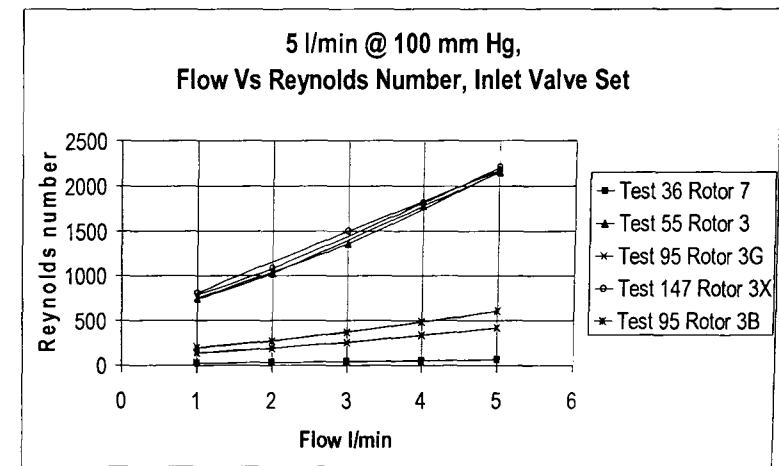
Appendix 14.76 – Flow Vs Flow parameter



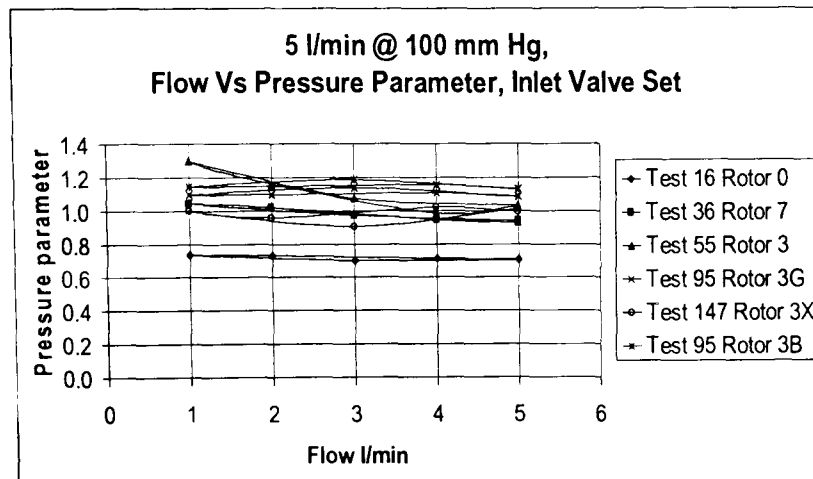
Appendix 14.77 – Flow Vs Reynolds number



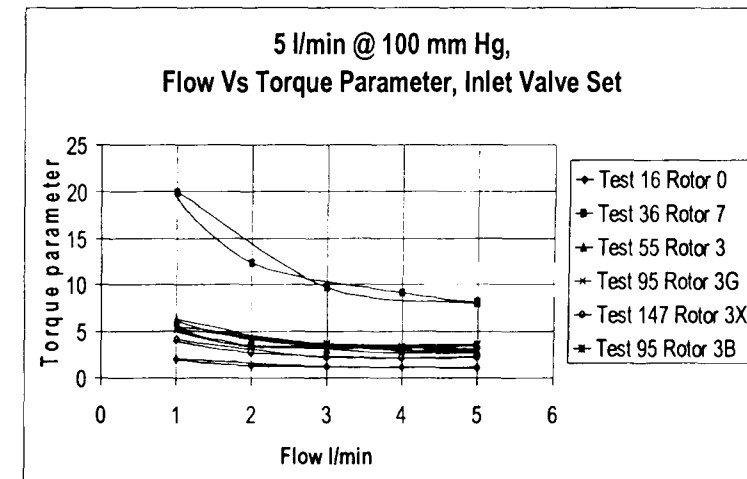
Appendix 14.78 – Flow Vs Reynolds number



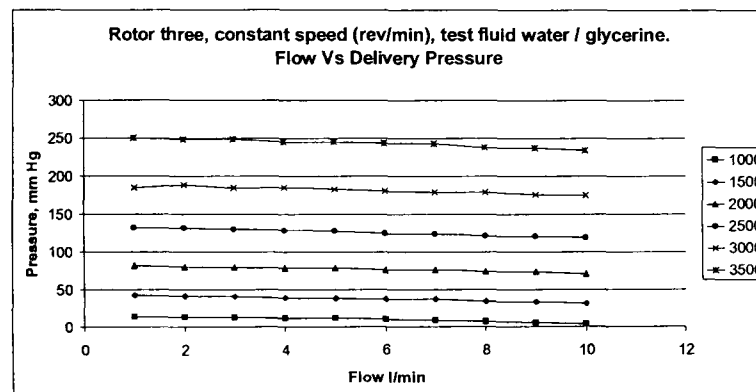
Appendix 14.79 – Flow Vs Pressure parameter



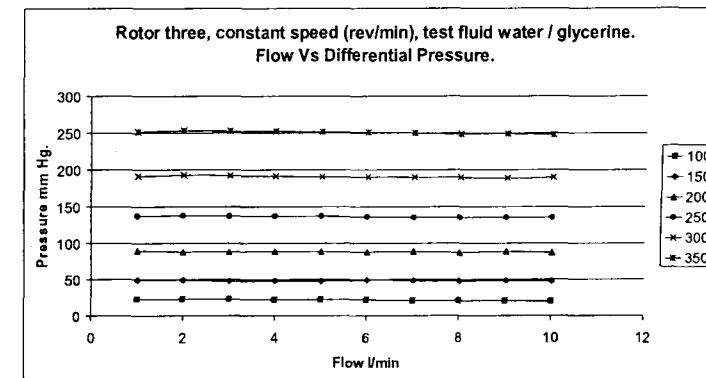
Appendix 14.80 – Flow Vs Torque parameter



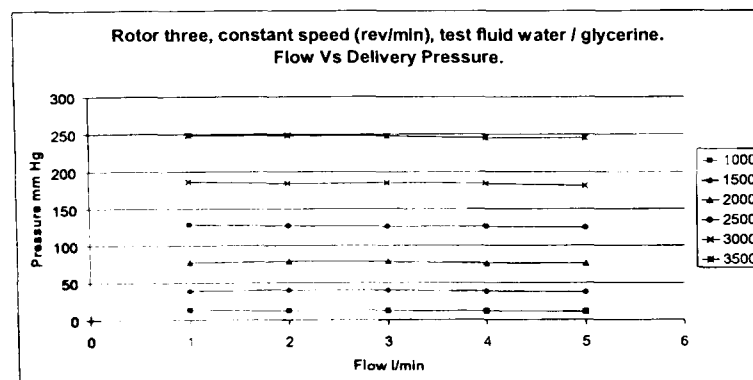
Appendix 14.81 Test fluid – water / glycerine. Constant speed tests, 1000 -3500 rev/min. Flow Vs Delivery Pressure - optimum rotor - flow of 10 l/min.



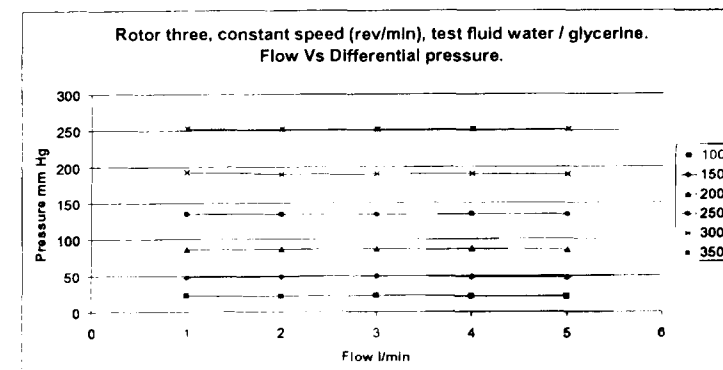
Appendix 14.82. Test fluid – water / glycerine. Constant speed tests, 1000 – 3500 rev/min. Flow Vs Differential Pressure - optimum rotor - flow of 10 l/min.



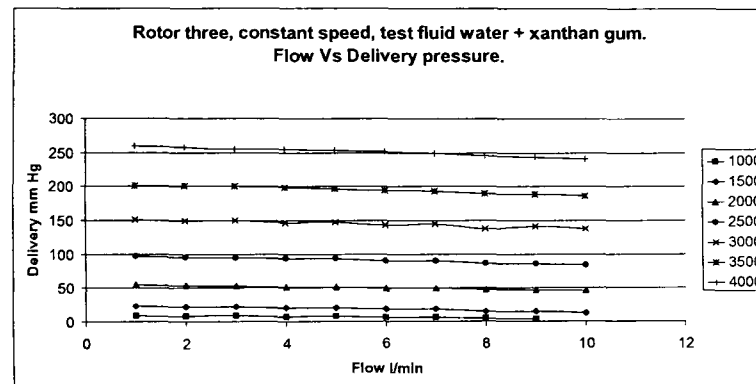
Appendix 14.83. Test fluid – water / glycerine. Constant speed tests 1000 – 3500 rev/min. Flow Vs Delivery Pressure - optimum rotor - flow of 5 l/min



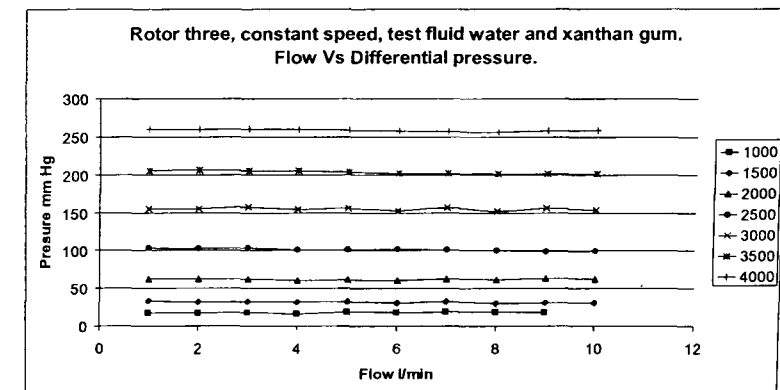
Appendix 14.84. Test fluid – water / glycerine. Constant speed tests, 1000 – 3500 rev/min. Flow Vs Differential Pressure - optimum rotor - flow of 5 l/min.



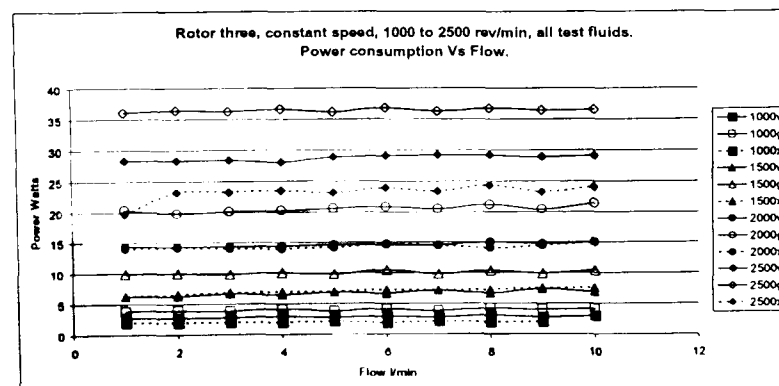
Appendix 14.85. Test fluid – water / xanthan gum.
Constant speed tests 1000 – 4000 rev/min. Flow Vs
Delivery Pressure - optimum rotor - flow of 10 l/min



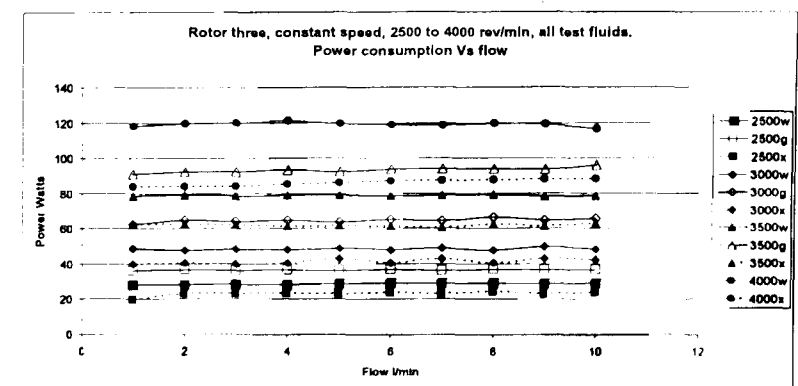
Appendix 14.86. Test fluid – water / xanthan
gum Constant speed tests 1000 – 4000 rev/min. Flow Vs
Differential Pressure - optimum rotor - flow of 10 l/min



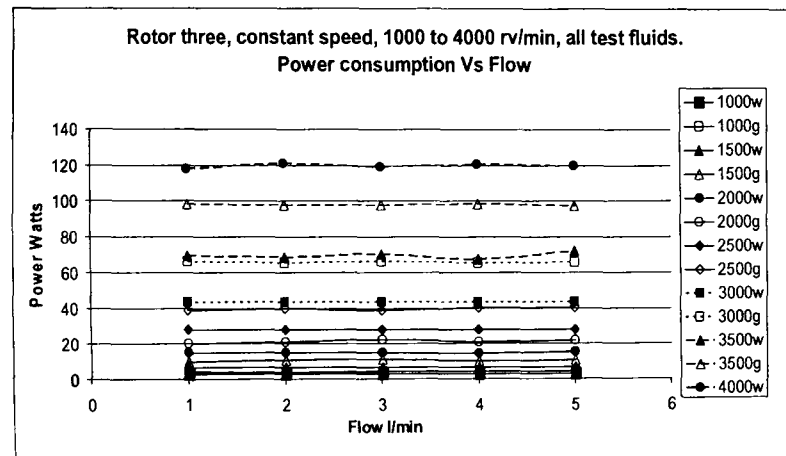
Appendix 14.87. All test fluids. Constant speed tests
Tests 1000 - 2500 rev/min. Flow Vs Power –optimum
rotor – flow of 10 l/min.



Appendix 14.88. All test fluids. Constant speed
tests 2500 – 4000 rev/min. Flow Vs Power – optimum
rotor -flow of 10 l/min.



Appendix 14.89. Two test fluids. Constant speed tests
 1000 – 4000 rev/min. Flow Vs Power – optimum
 rotor – flow of 5 l/min.



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